STUDY OF THE WEAR RATE AND IT'S EFFECTS IN CEMENTLESS TOTAL HIP ARTHROPLASTY

A dissertation submitted to the Tamil Nadu Dr.M.G.R. Medical University in partial fulfillment of the requirement for the award of M.S. Branch II (Orthopaedic Surgery) degree March 2008-2010

CERTIFICATE

This is to certify that this dissertation titled **"STUDY OF WEAR RATE AND IT'S EFFECTS IN CEMENTLESS TOTAL HIP ARTHROPLASTY"** is a bonafide work done by **Dr. ASOLIE CHASE,** in the Department of Orthopaedic Surgery, Christian Medial College and Hospital, Vellore in partial fulfillment of the rules and regulations of the Tamil Nadu Dr. M.G.R. Medical Unviersity for the award of M.S. Degree (Branch-II) Orthopaedic Surgery under the supervision and guidance of **Prof. VERNON N.LEE** during the period of his post-graduate study from March 2008 to Februrary 2010.

This consolidated report presented herein is based on bonafide cases, studied by the candidate himself.

Prof. Vernon N. Lee
D.Ortho., M.S.Ortho., M.Ch.Ortho
Professor of Orthopaedics
Department of Orthopaedic Surgery
Christian Medial College & Hospital
Vellore – 632004. TAMILNADU

CERTIFICATE

This is to certify that this dissertation titled **"STUDY OF WEAR RATE AND IT'S EFFECTS IN CEMENTLESS TOTAL HIP ARTHROPLASTY"** is a bonafide work done by **Dr. ASOLIE CHASE,** in the Department of Orthopaedic Surgery, Christian Medial College and Hospital, Vellore in partial fulfillment of the rules and regulations of the Tamil Nadu Dr. M.G.R. Medical Unviersity for the award of M.S. Degree (Branch-II) Orthopaedic Surgery under the supervision and guidance of **Prof. VERNON N.LEE** during the period of his post-graduate study from March 2008 to Februrary 2010.

This consolidated report presented herein is based on bonafide cases, studied by the candidate himself.

Prof. Vernon N. Lee
D.Ortho., M.S.Ortho., M.Ch.Ortho
Professor & Head,
Department of Orthopaedic Surgery
Christian Medial College & Hospital
Vellore.

ACKNOWLEDGEMENT

Firstly, I thank my Lord and Saviour Jesus Christ for His grace and constant guidance in my life.

I have great pleasure in thanking my teacher **Prof. Vernon N.Lee**, Professor and Head of the Department of Orthopaedic Surgery, Christian Medical College for permitting me to use the clinical materials and for guiding me through the preparation of the dissertation.

I owe my sincere thanks to **Dr. Manasseh Nithyanand**, Lecturer, Department of Orthopedic Surgery – Unit I for his guidance and encouragement throughout this study.

My sincere thanks to my teachers **Prof. G.D. Sunderaraj, Prof. Samuel Chittaranjan, Prof. Ravi Jacob Korula and Prof. Vrisha Madhuri** who gave valuable advise, support and encouragement throughout the preparation of this thesis.

I sincerely acknowledge the help rendered by **Ms Nithya Joseph**, Department of Biostatistics in performing the statistical analysis of data.

I am also grateful to other faculty members of the Department and my post-graduate colleagues who helped me in all possible ways in this study.

Grateful acknowledgement to Frank Cook,frank.cook@rjah.nhs.uk, Institute of Orthopaedics,Oswestry for permission to use Roman software.

INDEX

AIMS AND OBJECTIVES

The aims and objectives of this study are:

1. To assess the difference in polyethylene (PE) wear rate between conventional PE liner and cross linked PE liner in uncemented Total Hip Arthroplasty (THA).
2. To assess acetabular and femoral osteolysis in uncemented THA
3. To study the proximal femur stress shielding in uncemented THA.
4. To assess functional outcome in patients with conventional PE liner and cross linked PE liner.

INTRODUCTION

The Hip joint is a ball and socket type of synovial joint. It is a major weight-bearing joint and is subjected to high physiologic loads. Hence, it is one of the commonest joints in the human body to develop arthrosis worldwide. A Total Hip Replacement (THR) or Total Hip Arthroplasty (THA) is used in the treatment of an arthritic hip secondary to wear and tear, disease or injury that leads to painful destruction of the hip joint and hence, is a very commonly performed orthopaedic procedure worldwide.

A Total Hip Arthroplasty consists of a femoral stem, a femoral head, and an acetabular cup. The femoral stem is either of a modular or a monoblock design.

In the former, the femoral head is separate and is attached to the stem through a taper locking mechanism. In the latter, the femoral head and stem comes in one piece. Similarly, the acetabular cup is monoblock or modular; the monoblock cup is a one-piece construct whereas the modular cup consists of a shell that is fixed to the pelvic bone, and an insert (liner), which is fixed inside the shell. The bearing surface of the artificial joint thus is constituted by a metal or ceramic femoral head, and the inner surface of the cup, typically made of polyethylene (plastic), ceramics, or metal.

With the introduction of Low Friction Arthroplasty (LFA) by Sir John Charnley in 1962, there has been tremendous progress in the field of hip replacement

arthroplasty. Newer prosthetic designs have been developed with insight into the biomechanics of the artificial joint, newer bearing surfaces, newer surgical techniques and approaches, newer techniques in fixation of cemented and uncemented prosthesis.

Ultrahigh molecular weight polyethylene has been the preferred acetabular bearing material in THA for the past four decades. The fundamental limitation is the polyethylene wear and its associated complications.

This study aims to compare the wear rate in two types of polyethylene (conventional PE and cross linked PE) and assess its associated complications.

REVIEW OF LITERATURE

HISTORICAL REVIEW OF THA

1821 - Anthony White performed the first excision arthroplasty of the hip at the Westminster Hospital in London[1].

1860 - Auguste Stanislas Verneuil, Paris performed the first soft tissue Interposition [1]

1840 - Hip Arthroplasty: Carnochan replaced hip joint by wooden block[2].

1890 - Hemiarthroplasty: Gluck introduced ivory joint [1].

1919 - Hemiarthroplasty: Delbet used rubber in place of femoral head[1]

1925 - Hip Arthroplasty: M.N.Smith-Peterson,Boston,Massachusetts,USA, introduced Mould arthroplasty using glass [1] .

Another material was viscaloid (celluloid derivative).

1933 - Hip Arthroplasty: Material used was Pyrex[3].

1936 - Hip Arthroplasty: Material used was Vitallium (Co-Cr alloy).

1938 - Philip Wiles : First Total Hip Arthroplasty with a metal-on-metal prosthesis made of stainless steel[2]

1939 - Bohlman and Austin T.Moore used a 12-inch long Vitallium femoral head prosthesis in a patient with Giant Cell Tumour of the proximal femur[4].

1939 - Frederick R. Thompson of New York – Thompson prosthesis[2]

1946 - Judet brothers first develop the Acrylic short stemmed prosthesis[5].

1950s - Hemiarthroplasty: F. R. Thompson and Austin T. Moore used long stemmed prosthesis

1950 - Sven Kiaer introduced acrylic cement[2]

1952 - Gaenslen introduced metallic acetabular cup[2]

1955 - McBride introduced metallic acetabular cup used along with Thompson prosthesis[2]

1957 - Urist : Vitallium acetabular socket used along with Thompson femoral prosthesis[6]

1958 - Sir John Charnley develops Low Friction Arthroplasty (LFA) using Polytetrafluoroethylene (PTFE)[2]

1962 - Sir John Charnley - The first cemented metal-on-polyethylene hip replacement at the Wrightington Hospital in England using cemented high-density polyethylene (UHMWPE) socket and monoblock cemented femoral stem with head size of 22.225 mm[7].

1963 - McKee and Watson-Farrar - Metal-on-metal articulation with a modified Thompson femoral head prosthesis and a chrome-cobalt metal socket fixed with cement[8].

1964 - Ring prosthesis : Acetabular cup with a long threaded stem and a modified Moore's prosthesis as femoral stem[9]

1970 - Pierre Boutin first implanted alumina-on alumina ceramics[2]

1973-76 - McKee Farrar used CoCrMo alloy[2]

1980 - Silane cross-linked HDPE – Wrightington Hospital[2]

1990s - New COC (alumina and zirconia) Design

1992 - Sedel introduced COC (Alumina)[2]

1995 - Muller used cobalt chrome alloy pairings[2]

2000s - HAP: Interest on Hydroxyapatite materials with porous surface to encourage bone ingrowth.

EVOLUTION OF TEFLON AND POLYETHYLENE IN THA

The first known joint arthroplasties were made of ivory or platinum. Defining success as more than 50% good results, it took some 50 years to get there [2]. The mould hemi-arthroplasties of M. N. Smith-Petersen were made of glass, pyrex or bakelite the first fifteen years [3]. These failed because of material fragility, high friction, and foreign body reactions. From 1937, Smith-Petersen used Vitallium (CoCr), and this mould arthroplasty had sufficient mechanical strength to provide long durability.

In 1958, Sir John Charnley aggressively pursued effective methods of replacing both the femoral head and acetabulum of the hip and he developed a conceptual

low friction arthroplasty after analysing animal joint lubrication. He realized that a cartilage substitute was necessary in order to allow artificial joint to function at extremely low friction level as seen in nature. His first attempt was to use Teflon shells on the surface of the femoral head and acetabular components[2].

Teflon or polytetrafluoroethylene (PTFE) is a synthetic fluoropolymer of tetrafluoroethylene, which was accidentally invented by Roy Plunkett in 1938[10]. The coefficient of friction of PTFE is 0.1 or less, which is second lowest of any solid material ,after diamond-like carbon . Inspite of its low friction, it cannot be cross linked like an elastomer ,due to its inertness. Hence it is subject to creep and has low resistance to wear[11].

The rapid failure of Teflon parts led to development of a new design with a small diameter metallic femoral head attached to acrylic-fixed stem, which articulated with a thick walled Teflon shell .This new design failed quickly due to the poor wear characteristics, and led to generation of huge amount of wear debris. These wear debris promoted massive inflammatory reactions in the joints and travelled to various parts of the body via blood. Teflon was abandoned in 1962 because of serious tissue reactions, often associated with caseation and sterile pus formation[12].

This led to the development of a socket made of Ultra High Molecular Weight Polyethylene UHMWPE (trade name R.C.H. 1000) with wear properties that was 500 to 1000 times better than Teflon[12] . Ultra-high molecular weight polyethylene

(UHMWPE) is a unique polymer with outstanding physical and mechanical properties. It is significantly more abrasion and wear resistant than High Density Polyethylene (HDPE).

In 1962, Charnley finalized his total hip design - the cemented high-density polyethylene socket and the monoblock cemented femoral stem with head size of 22.225 mm. This was polished and manufactured out of EN58J stainless steel. There have been five changes in the standard 'Charnley' stem since these first-generation flat-back stems[13]. There have also been various other prosthetic designs evolved from the Charnley hip.

Since its introduction in 1962 by Charnley, UHMWPE articulating against a metallic femoral head remains the gold standard bearing surface combination for total joint arthroplasty. The 1970's saw the clinical introduction of a highly crosslinked UHMWPE by over 1000 kGy of gamma irradiation in air[14]. In the late 1980s, a joint venture between DePuy Orthopedics and DuPont developed a highly crystalline form of UHMWPE distributed under the trade name of Hylamer . The clinical history of Hylamer, which unfolded during the 1990s, has been mixed and therefore controversial[15].

Timeline of UHMWPE Development for Joint Replacement

1962 Charnley adopts UHMWPE for use in his LFA. Components were chemically sterilized.

1968 Start of Leeds production of the Charnley LFA by Chas F. Thackray, Ltd., of Leeds. The UHMWPE was gamma irradiated.

1969 General commercial release of the Charnley LFA by Chas F. Thackray, Ltd., of Leeds. UHMWPE were marketed as gamma irradiated (in air) with a minimum dose of 2.5 MRad.

1970s Commercial release of the Poly II--Carbon Fiber Reinforced . UHMWPE for THA/TKA by Zimmer, Inc

1972 Use of alumina ceramic heads articulating against UHMWPE in Japan.

1980-84 Co-development of Silane-Crosslinked HDPE by University of Leeds, Wrightington Hospital, and Thackray.

1980s Commercial release of Hylamer (Extended Chain Recrystalized UHMWPE) for THA/TKA/TSA by DePuy Orthopedics.

POLYETHYLENE WEAR AND COMPLICATIONS

The metal-on-polyethylene bearing couple remains the most common articulation 40 years after its introduction in total hip replacement (THR)[16].

The fundamental limitation with this bearing material is wear resistance. Wear is defined as the loss of material from a surface due to motion. Polyethylene wear and subsequent generation of polyethylene particles have been associated with osteolysis and subsequent loosening of prosthesis[18].

Wear particles can be generated by abrasive wear, in which particles are generated by rough articular surfaces (e.g. scratches, carbide asperites), either at the primary articulation or secondary surfaces (backside of PE insert with metal backed shell). Third-body particles can generate debris through an abrasive process at the articulating surfaces. McKellop[26] also describes the loss of PE at the primary articulation via a process of adhesive wear. PE is produced by the heating of small PE beads into a congealed mass. On its surface, the small submicron beads can be pulled off by the passing of the adjacent articulation surface. The combination of abrasive wear and adhesive wear can generate many billions of particles that are then disseminated through the effective joint space.

PE wear is related to several factors:

1. PE manufacturing

2 .Post processing sterilization

3. Shelf storage time

PE MANUFACTURING AND PROCESSING:

Polyethylene (PE) is a thermoplastic polymer consisting of long chains of the monomer ethylene. Polyethylene contains the chemical elements carbon and hydrogen.

Ultra high molecular weight polyethylene (UHMWPE) is a subset of polyethylene. It has extremely long chains, with molecular weight numbering in the millions (2 and 6 million). The high molecular weight makes it a very tough material, with the highest impact strength of any thermoplastic presently made. It is highly resistant to corrosive chemicals, with exception of oxidizing acids. It has extremely low moisture absorption, has a very low coefficient of friction, is self-lubricating, and is highly resistant to abrasion (15 times more resistant to abrasion than carbon steel). Its coefficient of friction is comparable to that of Teflon, but UHMWPE has better abrasion resistance than Teflon. It is odorless, tasteless, and nontoxic.

UHMWPE is produced as powder, the main ingredients being ethylene (a reactive gas), hydrogen and titanium tetra chloride (the catalyst). It is produced in 2 types, Type 1 and Type 2 resin with the trade names of GUR 1020 and 1050, respectively.

The UHMWPE powder is then consolidated into its solid form under elevated temperatures and pressures because of its high melt viscosity.

The UHMWPE for prosthetic devices in general is made by four different manufacturing techniques:

1. Ram bar extrusion with secondary machining into the desired product.
2. Hot isostatic pressing into large sheets with secondary machining into the desired product.
3. Compression molding into bars with secondary machining into the desired product.
4. Direct-compression molding from PE powder to the desired product.

One advantage of direct-compression molding is the extremely smooth surface finish obtained with a complete absence of machining marks at the articulating surface. In addition, higher processing pressures may be attained, if desired, because the projected surface area of each individual part mold is relatively small compared to the area of large molds used to compression mold sheets.

In addition, Calcium stearate was added as an additive by 1955. This additive acts as a scavenger for residual catalyst components that can potentially corrode conversion equipment. Calcium stearate also acts as a lubricant and a release agent. However, it was found that calcium stearate added to PE adversely affects the PE consolidation by creating areas with unfused PE particles (fusion defect).These

fusion defect areas significantly diminish the mechanical properties of finished PE implants. By 2002, the production of calcium stearate-containing resins (GUR 1120 and 1150) were discontinued.

There is currently no consensus as to which resin and conversion method would be universally superior for all orthopedic applications.

STERILISATION AND ITS EFFECT ON PE WEAR:

The conditions under which ultra-high molecular weight polyethylene cups are sterilized can markedly affect their long-term wear properties, and new sterilization methods and other modifications have been developed to minimize the negative effects.

Various sterilization methods:

1. Gamma sterilization in air:

Starting in the 1960s, UHMWPE components for joint replacement have been stored in air-permeable packaging and gamma sterilized with a nominal dose of 25 kGy (2.5 Mrad).

Air-permeable packaging was replaced by barrier packaging with a low oxygen environment starting during the mid 1990s. The reason for changing packaging techniques was to prevent oxidation of free radicals in the UHMWPE, which persist for years after irradiation and can be replenished by the cascade of chemical reactions that follows oxidation[18] . During shelf storage, UHMWPE components

that were gamma sterilized in air-permeable packaging undergo oxidative degradation, resulting in an increase in density and crystallinity, and more importantly, in a loss of mechanical properties, associated with progressive embrittlement[19].

This method has been discontinued by major orthopaedic manufacturers, but will likely continue as a clinically relevant issue as it was used till the mid 1990s.

2 .Gamma sterilization in barrier packaging:

This method consist of evacuating the air from the packaging and backfilling with an inert gas such as nitrogen or argon. The "barrier" in the package consists of polymer laminates or metallic foils to block gas diffusion. The goal of barrier packaging is to minimize oxidative degradation during long-term shelf storage.

Current studies have suggested that barrier packaging prevents oxidative degradation of UHMWPE during shelf storage even if it was originally sterilized in the presence of air[20].

3.Ethylene oxide gas sterilization:

Ethylene oxide (EtO) is a highly toxic gas which neutralizes bacteria, spores, and viruses. UHMWPE is a good candidate for EtO sterilization, because it contains no constituents that will react with or bind to the toxic gas. However, because of its toxicity and hazardous residues, ethylene oxide sterilization is conducted in accordance with domestic and international standards[21]. Laboratory studies suggest

that sterilization using ethylene oxide gas does not substantially influence the physical, chemical, and mechanical properties of UHMWPE[22].Based on a limited number of retrieval studies, the clinical experience with EtO-sterilized UHMWPE components has thus far been favorable[23].

4. Gas Plasma sterilization:

Low-temperature gas plasma is a relatively new commercially-available sterilization method that was applied to UHMWPE in the 1990s. Gas plasma is a surface sterilization method that relies upon ionized gas for deactivation of biological organisms .Gas plasma is an attractive sterilization method because it does not leave toxic residues or involve environmentally hazardous byproducts[24] . Recent laboratory investigations suggest that low temperature gas plasma does not substantially affect the physical, chemical, or mechanical properties of UHMWPE[24] . Because of its recent introduction, retrieval data from in vivo gas plasma sterilized UHMWPE components are not yet available.

The use of gas plasma or ethylene oxide sterilization methods generate no free radicals that can subsequently oxidize during shelf storage. However, UHMWPE sterilized in this manner also does not receive a tribological benefit associated with radiation-induced crosslinking. UHMWPE sterilized with low dose radiation (2.5-4.5 Mrad) in an inert environment without oxygen favors cross-linking of PE. Cross-linking improves adhesive and abrasive wear which in turn improves

bearing wear rates. UHMWPE treated with high dose irradiation is named highly cross-linked PE.

However, cross-linking also has disadvantages .Increasing cross-linking results in diminished mechanical properties of PE. Highly cross-linked PE has shown a significant reduction in Young's modulus, yield strength and fracture toughness. It also has a diminished fatigue-crack resistance.

McKellop et al.[25] reported on the wear performance of UHMWPE in a contemporary hip simulator following gamma irradiation in air, gamma irradiation in an inert gas, ethylene oxide, or gas plasma. Between 2 and 5 million cycles (each million of cycles corresponds to about a year of use in vivo for an average patient), the wear rate of the gamma sterilized UHMWPE was significantly lower than UHMWPE sterilized by either gas plasma or ethylene oxide. For example, the wear rate from 3.5 to 5 million cycles for ethylene oxide-sterilized UHMWPE was reported as 40 ± 0.6 mm^3/million cycles; in contrast, the wear rate for UHMWPE that was gamma irradiated in an air-permeable package was found to be 18.5 ± 0.9 mm^3/million cycles .When oxygen is excluded from the package during sterilization, further crosslinking, and additional improvement in wear performance, may be achieved relative to gamma sterilization in air[24].

Currently , there is no clear consensus on superiority of a particular sterilization method in terms of better clinical wear behavior.

SHELF LIFE OF UHMWPE COMPONENTS

PE performance can also be adversely affected by its shelf storage time (the time the PE product sits on the shelf before implantation). The extent of PE degradation during shelf storage time depends on two major factors:

1) Extent of irradiation,

2) Type of packaging.

Irradiation of PE produces free radicals[19]. If PE products are packaged and irradiated in an oxygen-free environment, oxidation of PE is minimized. However, if PE products are allowed to sit on the shelf and age, oxygen diffusion into PE can occur, resulting in oxidation. Free radicals are known to survive within PE for as long as 2-3 years .The higher the radiation dose, the more free radicals that are produced. Therefore, a long shelf life can adversely affect PE performance via on-the-shelf oxidation. Non irradiated PE products are not affected significantly by shelf storage time.

Until recently, a consensus practice of 5-year shelf life was adopted in Europe for medical implants so that sterility can be assured. However, within the past few years, the European Committee for Standardization (CEN) has established standards that limit the shelf life of UHMWPE components to 5 years.

OSTEOLYSIS

Osteolysis refers to an active resorption of bone matrix by osteoclast as part of an ongoing disease process or infection or inadequate blood supply.

In joint replacement arthroplasty, periprosthetic osteolysis refers to bone loss or appearance of endosteal cortical erosions along the component that was not identifiable on the immediate postoperative radiograph.

Periprosthetic osteolysis or aseptic loosening remains the most significant long-term complication with total hip replacement[26]. It has been reported with all materials and prosthetic devices in use or that have been used to date. Both the acetabular and femoral components may be affected.

HISTOLOGY

The formation of a ‘‘synovial-like membrane’’ between implant and bone is fundamental to most theories of aseptic loosening[27].Histological analysis of tissue surrounding loosened components after joint replacement reveals the presence of three distinct zones:

(1) a thin synovial layer of lining cells supported by fibrovascular tissue at both the cemented and bone surface;

(2) a middle layer containing histiocytes (tissue macrophages), giant cells, mononuclear cells (lymphocytes and mast cells) and periprosthetic particles; and

(3) a fibrous layer that blends into the marrow spaces between bone.

PATHOGENESIS OF BONE LOSS FOLLOWING TOTAL HIP REPLACEMENT:

Normal bone maintenance depends on the balance of bone formation and bone resorption that mainly involves the coordinated function of osteoblasts and osteoclasts. There are several mechanisms by which bone loss after a joint replacement may occur.

AGEING

Bone loss may occur as a result of natural ageing. Women can lose up to one third of their cortical bone and half of their trabecular bone throughout their lifetime, while men lose about 60% of that amount. However, bone loss secondary to the ageing process has not proved to represent a major threat to the mechanical stability of prosthetic components[28].

MECHANICAL FACTORS

Migration of prosthesis is defined as a change in position of prosthesis, cement mantle or both and is thought to indicate implant failure and represent loosening[29]. Once migration has begun, stability is lost and periprosthetic particles may modulate latter stages of loosening[31]. Mechanisms by which migration occurs are not fully understood. It could be due to fatigue failure of cancellous bone surrounding the prosthesis leading to loss of osteo-integration of a stable

prosthesis, or it could be attributed to surgical techniques—for example, reaming which disturbs capillary circulation of periprosthetic bone, leading to necrosis.

FLUID PRESSURE

Once a synovial-like membrane has formed, synovial fluid pressure within the joint may cause osteolysis[30]. With loading on the prosthesis, pressure on fluid within the membrane may rise significantly. Sustained elevated pressure can ultimately disturb normal perfusion and oxygenation of bone and, when transmitted to the membrane–bone interface, results in osteocyte destruction and bone necrosis.

PARTICULATE DEBRIS

It is now widely accepted that bone loss secondary to a biological reaction to particulate debris from implants is the principal mechanism responsible for periprosthetic osteolysis[31]. Particulate polyethylene is considered to be the substance causing the most tissue reaction, forming up to 90% of the debris volume[32]. Other particles that have been implicated in development of osteolysis include submicron-sized UHMWPE, polymethylmethacrylate (PMMA), and metallic debris such as cobalt and titanium alloys, silicates and stainless steel[34]. These particles probably exert their effects by either promoting third body wear of polyethylene, with UHMWPE triggering the cellular response; or they instigate the

release of inflammatory mediators which results in chronic inflammation and tissue damage that erodes the supporting bone with subsequent implant loosening.

MIGRATION OF PARTICLES AND THE CONCEPT OF "EFFECTIVE JOINT SPACE"

The concept of effective joint space , which includes all periprosthetic regions that are accessible to joint fluid and thus particulate debris, has been proposed as a mechanism for migration of particles[33]. Presence of particulate matter in joint fluid will initiate a localized macrophage-induced phagocytosis and result in bone resorption. As bone is resorbed, a pool is formed, promoting more flow (preferential flow) into that region and thus delivering more particles and causing more localized bone resorption[33]. This cycle continues and eventually a significant quantity of bone is resorbed which becomes evident as an osteolytic area on a radiograph. As fluid pressure propels joint fluid and thus particulate debris through the effective joint space, it will result in progressive bone loss[33].

Small particles (0.5–10 microm) are the most active and when generated will follow a route of least resistance and become interposed between the bone–cement interface or between the bone–implant interface in uncemented prostheses[34].

CELLULAR RESPONSE TO PARTICLES

The presence of particulate debris initiates phagocytosis by macrophages and macrophage-derived foreign body giant cells. As a consequence, macrophages and

possibly other cells including fibroblasts release cytokines such as tumour necrosis factor-a, interleukins (IL-1, IL-6, IL-10), proteolytic enzymes and prostaglandins (PGE2)[33]. Osteoblasts may also cause secretion of specific cytokines by activated macrophages. These intracellular mediators induce a complex cellular response, which initiates a focal bone resorptive process mediated primarily by osteoclasts and to a lesser degree by monocytes[33] .This in turn results in loosening of components.

BIOLOGICAL RESPONSE TO WEAR DEBRIS

Presence of wear debris does not always result in osteolysis. For osteolysis to occur, rate of production of wear particles must exceed an individual's capacity to remove the debris such that a threshold is reached above which development of osteolysis is more likely[35]. Furthermore, normal repair mechanisms that are responsible for preventing formation of osteolytic lesions must become unable to halt the disease progression[37].Therefore, an individual's biological response to presence of wear debris must play an important role in development of osteolysis. The rate of progression seems to be higher in patients with prosthetic loosening[28].

WEAR AND OSTEOLYSIS

Literature has shown that periprosthetic osteolysis and implant loosening is directly related to the wear of PE. Dumbleton et al.[37] surveyed the literature on wear and osteolysis around prosthetic hip implants and found that the appearance

of osteolysis increases as the rate of wear increases and that osteolysis is rarely observed in association with a wear rate of <0.1 mm/yr.

CLINICAL AND RADIOLOGRAPHIC MANIFESTATIONS OF OSTEOLYSIS

Radiolucent lines are seen around loose prosthesis on radiographs, most commonly in lateral and anterior aspects of the femur. Radioisotope scans may reveal areas of increased activity in areas of loosening. In the majority of cases, radiographic evidence of the disease process only manifests five years or more after insertion of the prosthesis[28]. Clinically, most patients are asymptomatic and diagnosed only following an incidental finding on late postoperative radiographs[28]. In a minority of cases, patients are symptomatic and present with thigh pain (usually indicates femoral component loosening), groin pain (usually indicates acetabular loosening) or fractures of the femur or acetabulum.

STRESS SHIELDING

Stress shielding refers to the reduction in bone density (osteopenia) as a result of removal of normal stress from the bone by an implant. This is explained by Wolff's law, which states that bone in a healthy person or animal will remodel in response to the loads it is placed under. Therefore, if the loading on a bone decreases, the bone will become less dense and weaker because there is no stimulus for continued remodeling that is required to maintain bone mass. Stress shielding, also called adaptive bone remodeling can occur in response to an altered mechanical

environment following a hip replacement. This occurs because there is a redistribution of load and therefore stress, when the femoral head is replaced by the femoral component of a total hip replacement. Consequently, stress on the proximal femoral cortex is lessened, as most of the load bypasses this area and is transmitted in the metal stem to the distal femur[36].

Interest in this phenomenon of stress-shielding led to the performance of numerous biomechanical and finite-element studies in an attempt to identify the factors that lead to bone-remodeling. These studies have led to the understanding that bone-remodeling is a ubiquitous phenomenon that occurs with all types of hip arthroplasty implants and is not unique to devices that are inserted without cement[37].However, cemented stems are associated with less stress shielding than uncemented stem[38].

Studies have shown that Hydroxyapatite fully coated stems are associated with an increased cortical bone stress shielding compared with proximally coated porous stems[39].The amount of coating on most prosthetic stems available today is still greater than that necessary to lower the stress-shielding effect on the proximal femur. However, reducing porous coating to lower stress shielding must be balanced against providing adequate coating to ensure fixation. Long-term effects of stress shielding on stability of components and further revision surgery are not known[33.]

It has been theorized that periprosthetic bone-remodeling secondary to stress shielding may contribute to increased pain or decreased function, fracture of the femur or the femoral component, loss of fixation of the implant, increased prevalence or severity of osteolysis, and difficulty in performing a revision[38].

Engh et al[39] in a long term study of 207 THAs found no adverse clinical consequences in the first ten year postoperative period in patients who had radiological evidence of stress shielding .

UNCEMENTED THA AND PE WEAR

Uncemented fixation regained popularity in the 1980's in the belief that the uncemented prostheses would provide better durability in the younger patients. This belief was based on reports of poor results of some cemented cups in young patients[40], and the belief that the bone cement itself was responsible for these poor results and for the periprosthetic osteolysis[41]. It is thought that the increased cycles and higher stresses applied to the hip joint in young, active patients lead to a more rapid failure of cemented components.

Principles of uncemented fixation include primary stability by press-fitting or screwing the components in the bone, and secondary fixation by ongrowth and ingrowth of bone to the implant surfaces. To achieve this, the implant surface is usually roughened by the means of blasting, porous coating, or Hydroxyapatite (HA) coating. HA stimulates bone growth adjacent to the implant.

The uncemented cup is typically a two-piece construct of a hemispheric metal backing and a polyethylene liner insert fixed to the shell by some kind of locking mechanism. Additional features of some cups include spikes, screws, fins, or pegs intended to provide additional fixation of the shell.

The uncemented femoral stem is typically a titanium modular stem. Various shapes and surfaces are commonly used, such as fit-to-fill in the proximal femur, rectangular cross-section, tapered or anatomical.

Uncemented THA is becoming increasingly popular and is now the preferred choice in younger patient categories. The results of uncemented total hip arthroplasty are varying. Generally, the durability in terms of implant fixation is good or excellent [41]. The problems of wear of the bearings and osteolysis however, suppress long-term implant survival [45]. Wear and osteolysis, which was blamed on the cement in the 1980's, are problems of even greater magnitude in uncemented hip arthroplasty. In order to address the problem of wear, newer bearings have developed during the last 20 years. Recent studies have reported better wear performance of these newer bearings as compared to the conventional bearing materials[43].

MEASUREMENT OF PE WEAR BY RADIOGRAPHIC METHODS

Radiographic methods for measuring polyethylene wear has evolved over the last thirty years from manual methods to a variety of computer-assisted techniques that can provide either two- dimensional or three-dimensional wear estimates. In addition, radiostereometric analysis has evolved and has been used successfully to measure femoral head penetration in vivo.

TWO-DIMENSIONAL MANUAL TECHNIQUES

All in vivo techniques estimate polyethylene wear on the basis of femoral head penetration relative to the acetabulum, with penetration of the head being assumed to represent the true loss of polyethylene material. The measurement of femoral head penetration cannot, however, differentiate bedding-in (consisting of creep of the polyethylene and/or settling of the liner) from the true loss of polyethylene material.

Charnley and Cupic[44] originally proposed a uniradiographic wear-measurement method that was used to determine the distance from the prosthetic femoral head contour to the contrast wire of the cup on the latest follow-up radiograph. Wear was calculated by subtracting the width of the narrowest measurement in the weight-bearing area from the width of the widest measurement in the non-weight-bearing area and dividing the difference by two. However, this technique did not

take magnification into consideration and it assumed that wear occurred mainly in the vertical direction.

Charnley and Halley[45] later introduced the duoradiographic technique, which used the same radiographic landmarks as seen on postoperative and follow-up radiographs. Wear was measured by subtracting the distance from the edge of the head to the contrast wire of the cup on the recent radiograph from the measured thickness of the same line on the initial radiograph after correcting for magnification. The accuracy of this technique was reported to be ±0.5 mm.

Scheier and Sandel modified the Charnley duoradiographic technique by locating the center of the femoral head with a template.

Livermore et al.[46] improved on these methods by using concentric circles on a template to locate the center of the femoral head and by using a compass to determine the location of the shortest radius from the center of the femoral head to a reference point on the acetabular cup. Wear was calculated as the difference between caliper measurements on the initial postoperative and follow-up radiographs. All measurements were corrected for magnification with use of the known diameter of the femoral head. The accuracy of this technique was determined by comparing radiographic measurements with direct measurements of the acetabular thickness of retrieved prostheses and initially was reported to be 0.075 mm (range, 0 to 0.4 mm).

Dorr and Wan[47] later described a uniradiographic method of measuring wear that could be used for metal-backed acetabular components. All measurements were corrected for magnification with use of the known diameter of the femoral head, but the authors assumed wear to be horizontal and did not report the accuracy of the technique.

Pollock et al.[48] described a uniradiographic technique that follows the dual-circle principle and involves the use of wear templates supplied by the manufacturer. The wear templates, which are created at 20% magnification (to match the magnification of the radiograph), depict a cross-sectional view of the cup and the thickness of the metal shell and show the original position of the femoral head. Wear is calculated by determining the remaining thickness of the polyethylene liner, which is accomplished by measuring the shortest distance between the edge of the femoral head and the inside of the metal shell. The authors who described this technique admitted that the measurements can be inaccurate by as much as 0.5 mm.

TWO-DIMENSIONAL COMPUTER-ASSISTED TECHNIQUES

In the 1990s, computer-assisted techniques were developed to reduce measurement variability and to more reliably measure femoral head penetration into the acetabular component. These techniques involved digitizing standard radiographs to create a computer model of the femoral head and the acetabular component.

Hardinge et al.[49] introduced the MAXIMA (Manchester X-ray Image Analysis) method of automatic image analysis. This was a duoradiographic method in which radiographs were digitized with a high-resolution camera, a copy stand, and a light box in order to increase the intensity, contrast, and consistency of points and lines. Reference lines were drawn interactively, and software was used to analyze changes in the position of the femoral head. This method was associated with high reproducibility, but no clinical studies were performed and no data on measurement accuracy were provided.

Ilchmann et al.[50] introduced the EBRA (Ein Bild Roentgen Analyze) method of wear measurement, which originally was designed for migration studies. This was a duoradiographic method that involved the use of a pencil, a ruler, and a digitizing table that was connected to a personal computer equipped with specially developed software. A grid of transverse and longitudinal tangents was drawn to define the position of the pelvis, and a simulated sphere was digitized on the basis of the gridlines. A comparability algorithm was then employed to divide the series of radiographs into comparable subgroups and to analyze the distance between gridlines. Wear-time diagrams were constructed in the horizontal and vertical directions with use of only comparable subgroups of radiographs. Although laborious, the EBRA method has shown high accuracy and has been used successfully in Europe for clinical studies[51].

Shaver et al.[52] developed an edge-detection technique that involved the use of digitized radiographic images. A software program was used to compute sampling rays emanating outward from the mathematically determined center of the femoral head. The edges of the acetabular and femoral components were identified with use of an edge-detection filter by evaluating the gradients of gray-scale intensity. After correction for magnification, femoral head penetration into the acetabular component was calculated with use of the dual-circle principle. The accuracy of this technique was evaluated in a series of laboratory benchtop studies and was reported to be 0.02 mm without supporting data.

THREE-DIMENSIONAL COMPUTER-ASSISTED TECHNIQUES

Devane et al.[53,54] described a three-dimensional measurement technique (PolyWare) for the measurement of polyethylene wear in metal-backed acetabular cups. This technique relied on computer-assisted technology to create a three-dimensional solid model of the acetabular component and femoral head on the basis of back projection of the radiographs (so-called shadow-casting) and CAD/CAM (computer-assisted design/ computer-assisted manufacturing) knowledge of the implant. With this technique, two-dimensional wear (in the frontal plane) was estimated on the basis of serial radiographs and three-dimensional wear was estimated by incorporating penetration as shown on lateral radiographs. In addition, an algorithm was used to estimate volumetric wear on the

basis of three-dimensional head penetration. In their initial article[56],Devane et al. used an acrylic phantom with a simulated head penetration of 8.55 mm and reported a three-dimensional accuracy of approximately 0.15 mm (on the basis of the mean absolute difference between the measured and true displacements) and a volume calculation that was within 8% of the true amount of the polyethylene removed. In addition, on the basis of multiple observations of one good-quality anteroposterior clinical radiograph and one good-quality lateral clinical radiograph, they reported an interobserver and intraobserver reproducibility of ±0.0768 and ±0.0493 mm, respectively (on the basis of the 95% confidence interval of the standard error). In 1999, Devane and Horne[55] reported improved reproducibility and accuracy in association with a more automated imaging protocol involving the use of a phantom setup consisting of two 38-mm-diameter steel balls.

Martell and Berdia[56] described a semi-automated computer-assisted dual-circle technique (Hip Analysis Suite [HAS]) that was based on edge detection and vector analysis of digital radiographs (so called shadow-comparing) for the determination of polyethylene wear in metal-backed acetabular components. This novel technique demonstrated approximately ten times better interobserver repeatability (a measure of precision) compared with the Livermore technique performed with either manual calipers or a digitizing tablet. In an analysis of fourteen retrieved acetabular liners, the wear estimates derived with use of the computer-assisted

technique differed by an average of 0.08 mm in comparison with the actual wear (as measured with use of an ultrasonic probe), which was substantially better than the estimates made with use of the Livermore technique. In addition, there was good agreement between the computer-assisted wear measurements and 2.0 mm of simulated wear (using a phantom setup in which Lucite was used to simulate soft tissue absorption and scatter effects). More recently, Martell et al.[57] reported on the use of this technique to provide three-dimensional wear data on penetration as seen on the lateral radiograph. The authors reported that three-dimensional analysis detected approximately 10% more wear than two-dimensional analysis did, but, because of the poor quality of the lateral radiographs, its repeatability was four times worse. They reasoned that the limited improvement in wear detection, coupled with the inferior repeatability, limits the usefulness of three-dimensional edge-detection techniques.

Geerdink et al[58] performed a study to compare the precision and usability of four computer-assisted methods in measuring linear wear rate (the Martell Hip Analysis suite 7.14, Rogan HyperOrtho, Rogan View Pro-X and Roman v1.70.).The intra and inter-observer variability for paired analysis was best for ViewPro-X and Roman software and worst for HyperOrtho and Martell. The Roman method proved the most precise and the most easy to use in clinical practice. The software is available free of charge[65].

RADIOISOMETRIC ANALYSIS (RSA)

In the early 1970s, Göran Selvik introduced roentgen stereophotogrammetric analysis, now commonly referred to as radiostereometric analysis (RSA)[59].Radiostereometric analysis is a highly accurate imaging technique that involves implanting tiny radiopaque (tantalum) beads in the human skeleton and around orthopaedic prostheses or hardware, thus allowing for the evaluation of three-dimensional micromotion. The measurement of polyethylene wear with use of radiostereometric analysis has been described for both metal-backed and non metal-backed components. For metal-backed acetabular components, tantalum markers are inserted into the polyethylene liner or attached to the end of specially designed towers that are locked into the metal shell. For non metal-backed components, markers usually are placed in the periacetabular bone or in the periphery of the component. Postoperatively, the patient is positioned over a specialized calibration cage and two simultaneous radiographs are made. The three-dimensional position of the femoral head with respect to the implanted beads can then be precisely determined over time with use of specialized computer software based on the cage coordinate system. The methodological details of radiostereometric analysis and corresponding software have been fully described[61]. Wear often is reported as proximal migration (vertical movement) and total migration (three-dimensional wear).

Several Swedish groups have done studies using radiostereometric analysis as a wear-measurement tool. There has been considerable variability in the reported wear rates as these studies have employed different radiostereometric analysis methods (such as placing beads in the polyethylene as opposed to into the periacetabular bone) and have examined a number of different implant designs and bearing surface materials.

Radiostereometric analysis (RSA) is currently the most accurate and precise method of evaluating motion of implants in vivo. Also the fact that few patients and short duration of time needed for in vivo wear assessment makes it an attractive alternative. However, this technique can be used only in prospective studies. RSA is still time consuming and demands trained personnel, special equipment and software, and economic resources. Hence, RSA still remains more of a research tool and is not practical for routine use.

ASSESSMENT OF THE ACCURACY AND PRECISION OF WEAR TECHNIQUES

Several investigations have been undertaken to explore the accuracy and precision of the various manual, computer-assisted, and radiostereometric analysis techniques.

In the study by Barrack et al.[60], wear estimates that were made with use of five different manual techniques and two computer-assisted versions of the Livermore technique were compared with wear measurements that were made with use of shadowgraph technique on twenty-one retrieved liners. The authors found a significant correlation between the radiographic and direct wear measurements with use of linear regression analysis ($p = 0.036$ to 0.00022); however, there was considerable variability between techniques. They concluded that radiographic wear measurements that are made with use of these techniques should be considered qualitative rather than quantitative. In addition, they thought that the addition of computer digitization to enhance manual methodology did not improve accuracy.

More recently, Hui et al.[61] reported that wear estimates that had been made with use of the Devane and Martell techniques were highly correlated with the actual measurements of two and three-dimensional linear and volumetric wear that were made with use of a coordinate-measuring machine for seventeen retrieved acetabular liners of a single design. The authors found some error or bias in association with both techniques (with PolyWare underestimating wear and the Hip Analysis Suite overestimating wear); the absolute difference between the radiographic estimates and the measured wear was approximately 19% (range, 13% to 24%).

Using both a phantom apparatus and retrieved acetabular liners, Ebramzadeh et al.[62] demonstrated that computerized wear methods (such as PolyWare and the Hip Analysis Suite) offered greater accuracy than a variety of manual methods did. However, the greatest improvement in accuracy was seen when the methods were used to evaluate laboratory radiographs (that is, radiographs of the hip phantom apparatus that were made in the laboratory); less improvement was observed when the methods were used to evaluate clinical radiographs.

With respect to accuracy and precision, Radiostereometric analysis has been repeatedly and widely validated with use of mathematical analyses, test-retest investigations and phantom studies. Bragdon et al.[63] performed a sophisti-

cated phantom study to evaluate the accuracy of radiostereometric analysis as a wear-measurement tool. Under ideal conditions (using beads attached to the femoral component), the accuracy was 0.033 mm for the medial direction, 0.022 mm for the superior direction, 0.086 mm for the posterior direction, and 0.055 mm for the resultant three-dimensional vector with corresponding precisions (at the 95% confidence level) of 0.0084, 0.0055, 0.016, and 0.0135 mm, respectively.

THE BEDDING-IN PHENOMENON

There is substantial evidence and general agreement that a considerable amount of the head penetration that occurs within the first years following the index

procedure represents the bedding-in phenomenon, a combination of settling of the modular liner and creep of the polyethylene. Under tensile load, UHMWPE will deform continually as long as the stress is present - an effect called creep.

In general, the steady-state (true) wear rate can be determined either retrospectively by plotting wear against time or prospectively by determining when the wear rate stabilizes (that is, when interval wear rates are not significantly different).To date, there is no clear standard for reporting wear with regard to defining a starting point or differentiating between steady-state wear and wear that includes bedding-in. Accurate and meaningful determination of the true rate of polyethylene wear may require starting wear analysis at twelve to twenty-four months postoperatively, after the majority of bedding-in has occurred.

MATERIALS AND METHODS

This study is a retrospective analysis of all primary uncemented Total Hip Arthroplasties done in the Department of Orthopaedics Unit 1, Christian Medical College, Vellore between September 2000 and December 2006.

Two group of patients who had uncemented THA with acetabular PE inserts of identical designs but different levels of crosslinking with a minimum follow of 27 months were compared.

EXCLUSION CRITERIA

1. Patients who had revision Total Hip Arthroplasty (THA).
2. Patients who had hybrid THA and cemented THA.
3. Patients who had THA after TB hip or post infection sequelae.

The study protocol was approved by our institutional review board (IRB).

This study was conducted between June 2008 and October 2009.

PATIENT DEMOGRAPHICS

GROUP 1: UNCEMENTED THA WITH CONVENTIONAL PE LINER

A total of 77 were patients were identified for the study in this group based on the operation theatre register, inpatient and outpatient records. This group included all patients who underwent primary uncemented THA with use of a conventional PE liner (Enduron; Depuy) as the acetabular bearing between Sept. 2000 to Oct. 2003.

Only 23 patients (29 hips) of the 77 (30%) were included in the final study. There were 19 men and 4 women, with a mean age of 37.4 years (range, 18-61 years) at the time of the index operation. The mean duration of follow-up was 71.83 months (range, 49 - 112 months).

The polyethylene liners used in this group were machined from non-cross-linked ram-extruded bars of GUR 1150 resin (Ticona, Summit, New Jersey) and were sterilized with use of gamma irradiation in air. All patients received the same cementless acetabular components (Duraloc; DePuy). The femoral heads were 28 mm in size and made of cobalt-chromium[64].

GROUP 2: UNCEMENTED THA WITH CROSS-LINKED PE LINER

A total of 102 were patients were identified for the study in this group. This group included all patients who underwent primary uncemented THA with use of a cross-linked PE liner (Marathon; Depuy) as the acetabular bearing between Oct. 2003 to Dec. 2006. Only 24 patients (27 hips) of the 102 (23.5%) were included in the final study. There were 21 men and 3 women, with a mean age of 42.9 years (range, 16-67 years) at the time of the index operation. The mean duration of follow-up was 42.3 months (range, 27 - 66 months).

The polyethylene liner used in this group was made of calcium-stearate-free GUR 1050 resin (Ticona). The cups were machined from the center of a ram-extruded bar that had been cross-linked by gamma irradiation to 5 Mrad and re-

melted at 155°C for twenty-four hours. The finished cups were sterilized with gas plasma. All patients received cementless modular acetabular components (Duraloc; DePuy). The femoral heads were 28 mm in size and made of cobalt-chromium[64].

The femoral component used in both groups was the AML stem (Anatomic medullary locking stem;DePuy,Warsaw,Indiana).This is a non modular stem with an extensive circumferential porous coating designed to be inserted without cement.

ASSESSMENT OF FUNCTIONAL OUTCOME

The functional outcome of patients in both groups was assessed using the Harris Hip Score (HHS). The domains included in the Harris Hip Score are

- Pain – 44 points
- Function – 47 points

  Gait – 33 points

  Activities of daily living (ADL) – 14 points

- Absence of deformity – 4 points
- Range of motion – 5 points

A score of 90-100 is considered as excellent, 80-89 good, 70-79 fair and below 70 poor.

9 patients with 10 hips in the Enduron group and 17 patients with 19 hips in the Marathon group who came for follow-up between June 2008 and October 2009 were examined and the Harris Hip Score proforma was filled with the pre-operative details made out from the hospital records.The pre-operative and post-operative Harris Hip Scores were analyzed and compared between the two groups. The pre and post-operative pain and function scores were also individually analyzed.

RADIOGRAPHIC ASSESSMENT

All patients in both groups have had anteroposterior and lateral radiographs taken on the third postoperative day after drain removal. They also had radiographs taken at the latest follow up visits.

The protocol followed for taking photograph after standard THA:

Patient positioning

For **AP view**

- Supine on the table with both the patellae pointing anteriorly (straight towards the ceiling). Rotate both legs medially or internally 15 degrees.
- Back against the film

- Anterosuperior Iliac spines at the same distance from the film.
- Central Ray (CR) should be perpendicular to the film.
- CR should be centered at the Pubic symphysis

CRITERIA FOR ASSESSING THE FILM

- Head of the femur and proximal half of femur bilaterally should be visible
- Lesser trochanter should be hidden.
- Median location of the symphysis and sacrum.

For **Lateral view**

- Lateral femoral surface against the film.
- Rotate femur medially or internally about 20 degrees.
- Move back the other femur.
- Central Ray should be perpendicular to the film and centered 10 cm distal to greater trochanter.

CRITERIA FOR ASSESSING THE FILM

- Acetabulum, hip joint and proximal half of femur should be visible.

The radiographs were retrieved for study from PACS using the GE Centricity software, Version 2.1.

RADIOGRAPHIC ASSESSMENT OF LINEAR WEAR OF PE

A computer-assisted method called Roman V1.70 was used to measure the linear penetration of the femoral head into the acetabular component.

Roman V1.70 (Robert Jones & Agnes Hunt Orthopaedic Hospital, Oswestry, United Kingdom)[65] is a radiological measurement program designed for orthopaedic application in general, and the linear penetration of the head can be measured by applying a digital equivalent of the manual Livermore method using the compass and ruler function of the software. Circles of best fit are drawn around the femoral head and the acetabular component after identifying a minimum of three points for interpolation on the contours of both components, and using the ruler function to measure the displacement of the centres of each circle.

The edges of the femoral head and the metal-backed acetabular component were determined manually by mouse-clicking on the edges. The distance between the centre of the femoral head and the centre of the acetabular shell can be measured after calibration of the image, using the known diameter of the femoral head to correct for magnification.

In the Livermore technique[46], a template with concentric circles was used to find the centre of the femoral head. The shortest distance from the femoral head centre to the outer surface of the cup defined the point of maximum wear, and this point was determined by the use of a compass. In the original method, a calliper is used to measure the thickness of the polyethylene at this point on the postoperative and

the latest follow-up radiographs. Linear wear, or rather femoral head penetration, is the difference found between the measurements. In this study, the method was used with slight modifications. Instead of measuring the polyethylene thickness directly, the distance from the centre of the femoral head to the outer surface of the metal backing was measured, assuming negligible wear of the metal femoral head.

The direction of wear was defined relative to a vertical line drawn through the center of the femoral head and perpendicular to a tangent to the ischial tuberosities. Measurement of the direction from the center of the femoral head to the thinnest portion of the acetabulum, if medial to this vertical line, was defined as a positive angle, θ; if the direction was lateral to this line, it was defined as a negative angle, α [46].

The volumetric wear was measured with the Livermore technique[46] by the formula $v = \pi r^2 w$, in which v is the volume of debris from wear (or volumetric wear), r is the radius of the femoral head, and w is the measured linear migration of the head through the polyethylene. This calculation assumes a uniform pattern of cylindrical wear.

Only the AP radiograph in JPEG image was measured for linear wear.

Each image was measured three times to minimize intra-observer variation.

OSTEOLYSIS

Osteolysis was defined as the appearance of areas of localized loss of trabecular bone or endosteal cortical erosion along the component that was not identifiable on the immediate postoperative radiograph[66].

Osteolysis around the acetabular component were recorded as being in one of the three zones described by DeLee and Charnley[67]. A component was defined as unstable if serial radiographs demonstrated a radiolucent line of more than one millimeter in width in all three zones.

Femoral osteolysis was recorded in terms of each of the seven zones on the anteroposterior radiograph, as described by Gruen et al.[68], and of a corresponding seven zones on the lateral radiographs. The severity of the osteolysis was described as 1) mild if the lesions occupied one or two zones; 2) intermediate if the lesions occupied three, four, or five zones; 3) extensive if the lesions occupied at least six zones[69].

STRESS SHIELDING (PROXIMAL FEMORAL RESORPTION)

Proximal femoral resorption or stress shielding was defined in the follow-up anteroposterior and lateral radiographs using the criteria described by Engh et al.[70]

The femur was divided into four levels on both AP and Lateral views. Each level was divided into medial, lateral, anterior and posterior sites to yield a total of 16 sites for examination.

The criterion for bone resorption in each site was whether the bone appeared darker, thinner or osteoporotic compared with immediately after surgery.

First degree - slight rounding of the proximal-medial edge of the cut femoral neck

Second degree - rounding of the proximal-medial aspect combined with loss of the medial cortical density to the level of the lesser trochanter (level 1)

Third degree - extensive resorption of cortical bone with involvement of the anterior cortex at the level of the lesser trochanter (level 1) and the medial cortex below the lesser trochanter (level 2)

Fourth degree - resorption extends into the diaphysis (below levels 1 and 2).

RESULTS

Statistical analysis of all data was done using the SPSS 11.0 for Windows software.

RADIOGRAPHIC FOLLOW-UP

CONVENTIONAL PE (ENDURON) GROUP

The mean duration during the last radiographic follow-up was 71.83 months with a range of 49-112 months.18 hips had radiographs with minimum of 5 years follow-up. The longest radiographic follow-up was 9 years and 4 months.

CROSS LINKED PE (MARATHON) GROUP

The mean duration during the last radiographic follow-up was 42.29 months with a range of 27-66 months. 16 hips had radiographs with more than 3 and a half years follow-up. The longest radiographic follow-up in this group was 5 years and 7 month.

LINEAR WEAR RATE

As the patient samples were heterogenous, statistical analysis was done using the Mann-Whitney test. Significance was defined as $p<0.05$.

The mean linear wear rate in the conventional PE group was 0.212 mm/yr (SD 0.122, range 0.058- 0.525) compared with 0.115 mm/yr (SD 0.139, range 0.02-0.75) in the X-linked PE group. This was statistically significant ($p<0.001$).

VOLUMETRIC WEAR RATE

The mean volumetric wear rate in the conventional PE group was 135.46 mm^3/yr (SD 79.05, range 18.11-323.26) compared with 55.55 mm^3/yr (SD 46.23, range 13.68-205.25) in the X-linked PE group. This was statistically significant ($p<0.001$). The volumetric wear in the X-linked group was 41 % lower than the conventional group.

DIRECTION OF WEAR

The direction of greatest wear was medial in 3 hips, lateral in 4 hips and neutral (caudally) in 22 hips in the conventional PE group.

The direction of greatest wear was medial in 9 hips, none in lateral and neutral (caudally) in 18 hips in the cross-linked PE group.

ACETABULAR OSTEOLYSIS

In the conventional PE group, 24 out of 29 hips did not have radiological features of acetabular osteolysis. 1 hip showed type 1 and 2 hips each showed type 2 and type 3 DeLee and Charley zones osteolysis.

All the 5 osteolytic lesions were >1.5mm in width. None of the lesions met the criteria for instability.

In the cross-linked PE group, none of the hips had radiological features of acetabular osteolysis.

This was statistically significant ($p<0.05$).

FEMORAL OSTEOLYSIS

In the conventional PE group, 19 out of 29 hips did not have radiographic features of femoral osteolysis. 7 hips had mild and 3 hips had intermediate osteolysis.

In the cross-linked PE group, 25 out of 27 hips did not have radiographic features of osteolysis. 1 hip each had mild and intermediate osteolysis respectively.

The difference in the occurrence of osteolysis was statistically significant ($p<0.05$).

The femoral stem in 1 patient in the conventional PE group had subsidence of >2mm in the latest follow up x-ray. However he did not have features of loosening clinically.

The femoral stem in 2 patients in the X-linked PE group had subsidence of >2mm in the latest follow up x-ray, but clinically did not have features of loosening.

None of the stems in both groups had a varus or valgus tilt >5 degrees.

STRESS SHIELDING

In the conventional PE group, 12 out of 29 hips did not have stress shielding on follow up x-ray. 8 had 1st degree, 8 had 2nd degree and 1 had 4th degree stress shielding respectively. None had 3rd degree stress shielding.

In the X-linked PE group, 11 out of 27 hips did not have stress shielding on x ray. 6 had 1st degree, 6 had 2nd degree, 3 had 3rd degree and 1 had 4th degree respectively. There was no statistically significant difference in the occurrence of stress shielding in both patient groups ($p>0.05$).

ANALYSIS OF FUNCTIONAL OUTCOME BY HARRIS HIP SCORE

9 patients with 10 hips in the conventional PE(Enduron) group were analysed for functional outcome. Preoperatively, the average Harris Hip score was 40.8 (range, 28 to 48). At the time of follow-up, the average Harris Hip score in this 85.6 points,with an average improvemnent of 45.4 points.

17 patients with 19 hips in the cross-linked PE (Marathon) group were analyzed for functional outcome. Preoperatively, the average Harris Hip score was 38.47 (range,

17 to 48). At the time of follow-up, the average Harris Hip score in this 85.47 points, with an average improvement of 47 points.

The improvement in the hip scores was statistically significant in both groups.

In the Enduron group, 3 had excellent and 7 had a good outcome.

In the Marathon group,3 hips had excellent, 12 had good and 3 had fair outcome.

The clinical results were further subanalysed to assess the degree of improvement of pain and function separately . Statistical analysis by the Levene's test and t-test showed no significant difference between the two groups with respect to improvement of pain, function or total Harris Hip Scores with a p value of > 0.05

COMPARISON OF WEAR RATE IN PATIENTS WITH OSTEOLYSIS AND WITHOUT OSTEOLYSIS

On combining both the groups, osteolysis was found in 15 (26%) of the 56 hips. Acetabular osteolysis was seen in 5 of the 56 hips. 12 hips had femoral osteolysis, and 2 hips had both acetabular and femoral osteolysis. Hips with osteolysis had higher linear and volumetric wear rates than hips without osteolysis.

DISCUSSION

Uncemented THA is a popular choice of hip replacement surgery, especially in younger age group patients. It offers excellent durability due to implant fixation by bone ingrowth and avoids potential loosening of cemented components in young active patients. However wear of the PE liner and debris mediated osteolysis remains a major concern. Cross linking of PE improves mechanical properties of the PE liner and hence improves the bearing wear rate. Hip simulator tests in laboratories have shown significant reduction in PE wear by cross linking, but the magnitude of reduction may not be as great in vivo. Also cross linking is known to produce much finer wear debris which may theoretically increase risk of osteolysis, thus affecting its clinical performance.

In this study, we have used a recognized radiological technique to measure the wear performance of 2 generations of PE (Conventional vs. cross linked PE) and also assess known complications of PE wear in an Indian population. The clinical performance of patients who underwent THA with these PE liners were also assessed and compared.

RADIOGRAPHIC RESULTS

WEAR RATE

The average linear wear rate in this study was 0.21mm/yr in the conventional PE group (Enduron) and 0.11mm/yr in the cross linked PE group (Marathon). This result was similar to those of Martell at al[71] which showed linear wear of 0.20 mm/yr in a group of 24 conventional PE versus 0.12 mm/yr in a group of 22 highly cross linked PE, with a follow up duration of 24 to 38 months. Dorr et al[72] showed wear rate of 0.32mm/yr in 37 conventional PE and 0.192 mm/yr in 37 highly cross linked PE at a follow up of 60 months. The wear rate on the conventional PE group in our study group was lower compared to those in Dorr's study.

Hooper at al[73] studied the wear in 48 Marathon PE and 50 Enduron PE at 24 months of follow up. The wear rate was 0.18 mm/yr in the Enduron PE and 0.08 mm/yr in the Marathon PE. The wear rate in the Enduron group in our study was slightly higher, but was much higher in the Marathon group as compared to the above study. One reason for the higher wear rate in this study could be the longer follow up period (Mean follow up in Enduron was 71.83 months and in Marathon was 42.29 months). Another reason may be the method used in measuring PE wear. We used a computer assisted software called the Roman V1.70 whereas Hooper used a computer-assisted radiographic measurement technique called the Sychterz method[75].

The mean volumetric wear rate was also correspondingly much lower in the Marathon PE as compared to the Enduron PE. There was a 41% reduction in wear rate in the Marathon PE (55.55 mm^3/yr) when compared with the Enduron PE (135.46 mm^3/yr). Heisel et a[74] measured the volumetric wear in 24 Enduron and 34 Marathon PE THAs with a minimum follow up period of 2 years and found 81% lower wear rate in the Marathon group.

Only one patient underwent PE liner exchange due to excessive PE wear and symptomatic groin pain. This patient had an uncemented THA 8 years and 9 months ago with an Enduron liner and had presented with groin pain of 3 months duration on the operated hip. The total linear wear measured was 4.2 mm (0.52mm/yr). Intermediate grade femoral osteolysis involving Gruen zones 1, 2, and 7 was also seen. There was no acetabular osteolysis.

ACETABULAR OSTEOLYSIS

In our study, 5 hips out of 29 hips (17.24%) in the Enduron group had features of osteolysis along the acetabular cup. 2 hips each had Dee Lee and Charley type 2 and type 3 osteolysis respectively and 1 hip had type 1 osteolysis . All the 5 osteolytic lesions were >1.5mm in width. None of the lesions met the criteria for instability (radiolucent line of more than one mm in all three zones) [75.] In the cross-linked PE group, none of the hips had radiological features of acetabular osteolysis. Zicat et al[75] reported a prevalence of 18% acetabular osteolysis in 74

uncemented THAs at mean duration of 105 months of follow up, out of which 3 cups required revision. Bitsch et al[76] ,in a 5 year follow up of 24 Marathon and 24 Enduron PE liner uncemented THAs, reported 2 patients with pelvic osteolysis in the Enduron group, 1 each in DeLee and Charnley zone 1 and 2 respectively. Osteolysis was not observed in any of the hips with a Marathon liner.

FEMORAL OSTEOLYSIS

In the Enduron group, 10 out of 29 hips (35%) showed radiological features of osteolysis on the latest follow up x-rays, of which 7 hips had mild and 3 hips had intermediate osteolysis. In the Marathon group, only 2 out of 29 hips (7%) had osteolysis on follow up x-rays. 1 hip each had mild and intermediate osteolysis respectively. None of these hips with osteolysis required revision due to aseptic loosening. None in either group had extensive grade of osteolysis.

Goetz et al[71] reported 29% (12 out of 41 hips) incidence of osteolysis in a study of uncemented THA with porous coated femoral and acetabular components with 6 years follow up. 6 hips had extensive osteolysis; 3, intermediate; and 3, mild. 5 femoral stems needed revision. The type of PE was not mentioned. Bitsch et al[76] reported 33% (8 out of 24 hips) incidence of femoral osteolysis in Enduron as compared to 0%(0 out of 32 hips) in Marathon PE THAs. This result was similar to ours. Engh et al[77] also showed comparable results in a similar study. Among 114 Enduron THA patients with minimum 4-year follow-up x-rays, 21% (19 out of 90)

had only femoral osteolysis, 19% (17out 90) had only acetabular osteolysis, and 18% (16 out of 90) had both acetabular and femoral lesions. Among 116 Marathon THA patients with minimum 4-year follow-up x-rays, 2% (2 out of 96) had only femoral osteolysis, 19% (18 out of 96) had only acetabular osteolysis, and 3% (3out of 96) had both femoral and acetabular lesions.

The higher incidence of osteolysis in the Enduron group in our study could possibly be a direct effect of higher PE wear rate. The patients in this group also had a longer follow up than the Marathon group which may also have resulted in the higher incidence of osteolysis.

Most of the osteolytic lesions occurred in the proximal segment of the femoral stem (Gruen zones 1 and 7) which supports the suggestion that extensive porous coating of the femoral stem prevents the distal migration of particulate debris, thus limiting the effective joint space.

STRESS SHIELDING

In our study, 96% of Enduron and 85% of Marathon hips had no changes or first and second degree changes, which according to Engh et al[70], are equivalent to little or no stress shielding .The similar incidence of proximal femoral stress shielding could possibly be due to similar type of femoral stem used in both groups ,the extensively porous coated AML stem, and also to the fact that most of the stress shielding occurs in the first 2 years after THA, as our patients had a longer

follow up. The high incidence of stress shielding in both the groups, although asymptomatic, agrees with literature that bone resorption occurs with porous coated femoral stems[72]. Although first and second degree stress shielding does not affect clinical results adversely, it would be ideal to have no bone resorption at all.

CLINICAL RESULTS

The clinical results of both groups in our study are in the accepted range for successful outcome after THA. The Harris Hip Score improved from 40.8 preoperatively to 85.6 postoperatively with an improvement of 44.8 points in the Enduron group, and from 38.4 preoperatively to 85.4 postoperatively with an improvement of 46.9 points in the Marathon group. There was no significant difference between the two groups.

Comparison with the Harris hip scores in several published series using contemporary uncemented THAs is favorable.

Kim et al[78] showed that the average preoperative Harris Hip Score in his study of porous coated uncemented THAs was 53.3 and that at the last follow-up visit was 91. McLaughlin et al[79] in a similar long term study on uncemented THAs showed the mean Harris Hip Score improved from 49.3 points preoperatively to 85.4 points postoperatively at the time of latest follow-up. This was comparable with our results.

Many of the patients in our study had inflammatory arthritis, 11(48%) patients in the Enduron group and 8(33%) patients in the Marathon group. This could explain for the slightly lower post operative mean score of 85 points as compared to other studies done in Western population where the main indication for THA is osteoarthritis.

CONCLUSION

1. Uncemented Total Hip Arthroplasty with cross linked polyethylene gives good long term clinical and radiological results.

2. The linear wear rate was 52% lower in THAs using cross linked PE liner as compared to those using conventional PE liners.

3. The incidence of femoral and acetabular osteolysis was lower between cross linked and conventional PE liner for uncemented THAs.

4. In this series, significant stress shielding was seen in only 19 % of patients.

LIMITATIONS OF THE STUDY

1. This is a retrospective study and offers level 3 evidence.

2. The final number of patients assessed in the study is only a small portion of the total number of patients who underwent uncemented THA during the study period, and may not be representative of the entire group.

3. The duration of follow up was different in the 2 patient groups studied and hence may contribute to non comparable radiological assessment.

4. Baseline characteristics such as Body Mass Index (BMI) and patient's activity level were not compared which may have a bearing on the wear rate of polyethylene.

5. The acetabular and femoral component malposition which can affect polyethylene wear was not assesed.

6. The method used for measuring the PE wear rate is subject to human error.

BIBLIOGRAPHY

1. Gomez P F, Morcuende J A. Early attempts at hip arthroplasty-1700s to 1950s. The Iowa Orthopaedic Journal. 25:25-29.

2. Pr amanik S, Agarwal A K, Rai K N. Chronology of total hip joint replacement and materials development.Trends Biomater Artif Organs. 2005; 19(1):15-26.

3. M.N. Smith-Petersen. Evolution of the Mould Arthroplasty of the Hip Joint. J Bone Joint Surg (Br) 1948; 30 B:59-75.

4. Moore A T, Bohlman H R. Metal hip joint: a case report. J Bone Joint Surg. 1943; 25:688.

5. Judet J, Judet R. The use of an artificial femoral head for arthroplasty of the hip joint. J Bone Joint Surg Br. 1950; 32-B:166-73.

6. Breck L W: Metal-to-metal total hip replacement using the Urist socket. An end result study. Clin Orthop. 1973; 95:38–42.

7. Charnley J. Total hip replacement by low-friction arthroplasty. Clin Orthop. 1970; 72:7-21.

8. McKee G K, Watson-Farrar J. Replacement of arthritic hips by McKee-Farrar prosthesis. J Bone Joint Surg Br. 1966 May; 48-B:245-59

9. Ring PA, Complete replacement arthroplasty of the hip by the ring prosthesis. J Bone Joint Surg Br. 1968;50:720-31.

10. Plunkett, Roy J, "Tetrafluoroethylene polymers", US 2230654, issued 1941-02-04

11. Fluoropolymer Comparison - Typical Properties www2.dupont.com. Retrieved 10 September 2006

12. John Charnley . Letter to the editor. J Bone Joint Surg Am.1966; 48:819.

13. Shen G. Femoral stem fixation: an engineering interpretation of the long-term outcome of Charnley and Exeter stems. J Bone Joint Surg Br. 1998;80- B:754-6.

14. Oonishi H. Long term clinical results of THR. Clinical results of THR of an alumina head with a cross-linked UHMWPE cup. Orthopaedic Surgery and Traumatology 1995; 38: 1255-1264

15. Kurtz S .M, Muratoglu O.K, Evans M, Edidin A.A. Advances in the processing, sterilization, and crosslinking of ultra- high molecular weight polyethylene for total joint arthroplasty. Biomaterials 1999; 20: 1659-88.

16. Callaghan J.J, Templeton J.E, Liu SS, et al. Results of Charnley total hip arthroplasty at a minimum of thirty years: a concise follow-up of a previous report. J Bone Joint Surg [Am] 2004;86-A:690-5.

17. Premnath V, Harris W H, Jasty M, Merrill E. W. Gamma sterilization of UHMWPE articular implants: an analysis of the oxidation problem. Biomaterials 1996; 17: 1741-1753.

18. Rimnac C.M, Klein RW, Betts F, Wright T.M. Post-irradiation aging of ultra-high molecular weight polyethylene. Journal of Bone & Joint Surgery 1994; 76A: 1052-1056.

19. Edidin A.A, Herr MP, Villarraga M.L, Muth J, Yau S.S, Kurtz S.M. Accelerated aging studies of UHMWPE. I. Effect of resin, processing, and

radiation environment on resistance to mechanical degradation. J Biomed Mater Res 2002; 61: 312-22.

20. Ries M.D, Weaver K, Beals N. Safety and efficacy of ethylene oxide sterilized polyethylene in total knee arthroplasty. Clin Orthop 1996; 331: 159-63. Collier JP, Sutula L.C, Currier B.H, Currier J.H, Wooding R.E, Williams I.R, Farber K.B, Mayor M.B. Overview of polyethylene as a bearing material: comparison of sterilization methods. Clin Orthop 1996; 333: 76-86.

21. Sutula L.C, Collier J.P, Saum K.A, Currier B.H, Currier J.H, Sanford W.M, Mayor M.B, Wooding RE, Sperling DK, Williams IR, et al. Impact of gamma sterilization on clinical performance of polyethylene in the hip. Clin Orthop 1995; 319: 28-40.24.Kyi MS, Holton J, Ridgway GL. Assessment of the efficacy of a low temperature hydrogen peroxide gas plasma sterilization 1system. J Hosp Infect 1995; 31: 275-84

24. McKellop H, Shen F.W, Lu B, Campbell P, Salovey R. Effect of sterilization method and other modifications on the wear resistance of acetabular cups made of ultra-high molecular weight polyethylene. A hip-simulator study. J Bone Joint Surg Am 2000; 82-A: 1708-25.

25. McKellop H.A, Shen F.W, Campbell P, Ota T. Effect of molecular weight, calcium stearate, and sterilization methods on the wear of ultra-high molecular weight polyethylene acetabular cups in a hip simulator. J Orthop Res 1999; 17: 329-339

26. Harris W.H. Wear and periprosthetic osteolysis: the problem. Clin Orthop 2001;393:66–70.

27. Goldring SR, Schiller A.L, Roelke M, et al. The synovial-like membrane at the bone-cement interface in loose total hip replacements and its proposed role in bone lysis. J Bone Joint Surg Am 1983;65:575–84.

28 . Jasty M, MaloneyW.J, Bragdon C.R, et al. Histomorphological studies of the long-term skeletal responses to well fixed cemented femoral components. J Bone Joint Surg Am 1990;72:1220–9.

29. Harris W.H, McCarthy J.C Jr, O'Neill D.A. Femoral component loosening using contemporary techniques of femoral cement fixation. J Bone Joint Surg Am 1982;64:1063–7.

30. Aspenberg P, Van d, V. Migration, particles, and fluid pressure. A discussion of causes of prosthetic loosening. Clin Orthop 1998;352:75–80.

31. Rubash H.E, Sinha RK, Shanbhag A.S, et al. Pathogenesis of bone loss after total hip arthroplasty. Orthop Clin North Am 1998;29:173–86.

32. Maloney W.J, Smith R.L, Schmalzried T.P, et al. Isolation and characterization of wear particles generated in patients who have had failure of a hip arthroplasty without cement. J Bone Joint Surg Am 1995;77:1301–10

33.Schmalzried TP, Jasty M, Harris W.H. Periprosthetic bone loss in total hip arthroplasty. Polyethylene wear debris and the concept of the effective joint space. J Bone Joint Surg Am 1992;74:849–63.

34. Jacobs JJ, Roebuck K.A, Archibeck M, et al. Osteolysis: basic science. Clin Orthop 2001;393:71–7.

35. Dumbleton J.H, Manley MT, Edidin A.A. A literature review of the association between wear rate and osteolysis in total hip arthroplasty. J Arthroplasty 2002;17:649–61.

36. McCarthy, C. K.; Steinberg, G. G.; Agren, M.; Leahey, D.; Wyman, E.; and Baran, D. T.: Quantifying bone loss from the proximal femur after total hip arthroplasty. J. Bone and Joint Surg. 1991; 73-B(5):774-778.

37. Maloney, W. J.; Sychterz, C. J.; Bragdon, C.; McGovern, T.; Jasty, M.; Engh, C. A.; and Harris, W. H.: Skeletal response to well-fixed femoral components inserted with and without cement. Clin. Orthop. 1996; 333:15-26

38. Jasty M, MaloneyWJ, Bragdon C.R, et al. Histomorphological studies of the long term skeletal responses to well fixed cemented femoral components. J Bone Joint Surg Am 1990;72:1220–9.

39. Bugbee, W. D.; Culpepper, W. J., II; Engh, C. A., Jr.; and Engh, C. A., Sr.: Long-term clinical consequences of stress-shielding after total hip arthroplasty without cement. J. Bone and Joint Surg., 79-A: 1007-1012, July 1997.

40. Chandler HP, Reineck F.T, Wixson R.L, McCarthy JC. Total hip replacement in patients younger than thirty years old. A five-year follow-up study. J Bone Joint Surg Am 1981;63(9):1426-34.

41 . Jones L.C, Hungerford D.S. Cement disease. Clin.Orthop.Relat Res. 1987(225):192-206.

42. Della Valle C.J, Berger R.A, Shott S, Rosenberg A.G, Jacobs J.J, Quigley L et al. Primary total hip arthroplasty with a porous-coated acetabular component. A

concise follow-up of a previous report. J Bone Joint Surg Am 2004;86-A(6):1217 -22.

43. D. Charron ,Richard W. McCalden, Steven J. MacDonald, Cecil H. Rorabeck, Robert B. Bourne, David G. Chess and Kory. Wear Rate of Highly Cross-Linked Polyethylene in Total Hip Arthroplasty. A Randomized control trial. J Bone Joint Surg Am. 2009;91:773-782.

44. Charnley J, Cupic Z. The nine and ten year results of the low-friction arthroplasty of the hip. Clin Orthop Relat Res. 1973;95:9-25.

45. Charnley J, Halley D.K. Rate of wear in total hip replacement. Clin Orthop Relat Res. 1975;112:170-9.

46. Livermore J, Ilstrup D, Morrey B. Effect of femoral head size on wear of the polyethylene acetabular component. J Bone Joint Surg Am. 1990;72:518-28.

47. Dorr L.D, Wan Z. Ten years of experience with porous acetabular components for revision surgery. Clin Orthop Relat Res. 1995;319:191-200.

48. Pollock D, Sychterz C.J, Engh C.A. A clinically practical method of manually assessing polyethylene liner thickness. J Bone Joint Surg Am. 2001;83:1803-9

49. Hardinge K, Porter M.L, Jones PR, Hukins D.W, Taylor CJ. Measurement of hip prostheses using image analysis. The maxima hip technique. J Bone Joint Surg Br. 1991;73:724-8.

50. Ilchmann T, Mjoberg B, Wingstrand H. Measurement accuracy in acetabular cup wear. Three retrospective methods compared with Roentgen stereophotogrammetry. J Arthroplasty. 1995;10:636-42.

51. Krismer M, Stockl B, Fischer M, Bauer R, Mayrhofer P , Ogon M. Early migration predicts late aseptic failure of hip sockets. J Bone Joint Surg Br. 1996;78:422-6.

52. Shaver SM, Brown T.D, Hillis SL, Callaghan JJ. Digital edge-detection measurement of polyethylene wear after total hip arthroplasty. J Bone Joint Surg Am. 1997;79:690-700.

53. Devane P.A, Bourne RB, Rorabeck C.H, Hardie RM, Horne JG. Measurement of polyethylene wear in metal-backed acetabular cups. I. Three-dimensional technique. Clin Orthop Relat Res. 1995;319:303-16

54. Devane P.A, Bourne R.B, Rorabeck C.H, MacDonald S, Robinson E.J. Measurement of polyethylene wear in metal-backed acetabular cups. II. Clinical application. Clin Orthop Relat Res. 1995;319:317-26.

55. Devane P.A, Horne J.G. Assessment of polyethylene wear in total hip replacement. Clin Orthop Relat Res. 1999;369:59-72

56. Martell J.M, Berdia S. Determination of polyethylene wear in total hip replacements with use of digital radiographs. J Bone Joint Surg Am. 1997;79:1635-41.

57. Martell J.M, Berkson E, Berger R, Jacobs J. Comparison of two and three-dimensional computerized polyethylene wear analysis after total hip arthroplasty. J Bone Joint Surg Am. 2003;85:1111-7

58. C. H. Geerdink, B. Grimm, W. Vencken, I. C. Heyligers, A. J. Tonino. The determination of linear and angular penetration of the femoral head into the acetabular component as an assessment of wear in total hip replacement. J Bone Joint Surg [Br] 2008;90-B:839-46.

59. Selvik G. Roentgen stereophotogrammetry. A method for the study of the kinematics of the skeletal system. Acta Orthop Scand Suppl. 1989;232:1-51.

60. Barrack R.L, Lavernia C, Szuszczewicz ES, Sawhney J. Radiographic wear measurements in a cementless metal-backed modular cobalt-chromium acetabular component. J Arthroplasty. 2001;16:820-8.

61. Hui A.J, McCalden R.W, Martell J.M, MacDonald S.J, Bourne R.B, Rorabeck C.H. Validation of two and three-dimensional radiographic techniques for measuring polyethylene wear after total hip arthroplasty. J Bone Joint Surg Am. 2003;85:505-11.

62. Ebramzadeh E, Sangiorgio S.N, Lattuada F, Kang JS, Chiesa R, McKellop H.A, Dorr LD. Accuracy of measurement of polyethylene wear with use of radiographs of total hip replacements. J Bone Joint Surg Am. 2003;85:2378-84

63.Bragdon CR, Malchau H, Yuan X, Perinchief R, Kärrholm J, Börlin N, Estok DM, Harris WH. Experimental assessment of precision and accuracy of radio-stereometric analysis for the determination of polyethylene wear in a total hip replacement model. J Orthop Res. 2002;20:688-95.

64. Christian Heisel, Mauricio Silva, Mylene A. dela Rosa and Thomas P.Schmalzried . Short-Term in Vivo Wear of Cross-Linked Polyethylene. J Bone Joint Surg Am. 2004;86:748-751.

65. No authors listed. Roman free to share software version V1.70. http:// www.Keele.ac.uk/depts/rjah/ .

66. M Tanzer, W.J Maloney, M Jasty and W.H Harris. The progression of femoral cortical osteolysis in association with total hip arthroplasty without cement. 1992;74:404-410. J Bone Joint Surg Am.

67. DeLee, J. G., and Charnley, J.: Radiological demarcation of cemented sockets in total hip replacement. Clin. Orthiop.. 121: 20-32. 1976.

68. Gruen, T. A.; McNeice, G. M.; and Amstutz, H. C.: "Modes of failure of cemented stem-type femoral components. A radiographic analysis of loosening. Cliii. Orthop., 141: 17-27. 1979.

69. D.D Goetz, E.J Smith and W.H Harris . The Prevalence of Femoral Osteolysis Associated with Components Inserted with or without Cement in Total Hip Replacements. J Bone Joint Surg Am. 1994;76:1121-1129.

70. Engh C.A, Bobyn J.D, Glassman A.H. Porous-coated hip replacement: the factors governing bone ingrowth, stress shielding, and clinical results. J Bone Joint Surg Br. 1987;69-B:45-55.

71. . Martell J.M, Verner J.J, Incavo S.J. Clinical performance of a highly cross – linked polyethylene at two years in total hip arthroplasty: a randomized prospective trial. J Arthroplasty 2003;18(Suppl 1):55-9.

72. Dorr L.D, Wan Z, Shahrdar C, et al. Clinical performance of a Durasul highly cross-linked polyethylene acetabular liner for total hip arthroplasty at five years.J Bone Joint Surg [Am] 2005;87-A:1816-21.

73. Hopper R.H, Young A.M, Orishimo K.F, McAuley J.P. Correlation between early and late wear rates in total hip arthroplasty with application to the

performance of marathon cross-linked polyethylene liners. J Arthroplasty 2003;18(Suppl 1):60-7.

74. Christian Heisel, Mauricio Silva, Mylene A. dela Rosa and Thomas P. Schmalzried . Short-Term in Vivo Wear of Cross-Linked Polyethylene. J Bone Joint Surg Am. 2004;86:748-751.

75. B Zicat, C,A Engh and E Gokcen .Patterns of osteolysis around total hip components inserted with and without cement. . J Bone Joint Surg Am. 1995;77:432-439

76. Rudi G. Bitsch, MD, Travis Loidolt, BA, Christian Heisel, MD, Scott Ball, MD, and Thomas P. Schmalzried, MD. Reduction of Osteolysis with Use of Marathon cross-Linked Polyethylene.J Bone Joint Surg Am. 2008;90:1487-91

77. C. Anderson Engh, Jr MD, Adam S. Stepniewski MD, Stuart D. Ginn BS, Sarah E. Beykirch BS, Christi J. Sychterz-Terefenko MS, Robert H. Hopper, Jr. PhD._A Randomized Prospective Evaluation of Outcomes After Total Hip Arthroplasty Using Cross-linked Marathon and Non–cross-linked Enduron Polyethylene Liners. The Journal of Arthroplasty, Volume 21, Issue 6, Supplement 1, September 2006, Pages 17-25

78. Young-Hoo Kim,Vana E.M.Kim.Uncemented porous-coated anatomic hip replacement.J Bone ioint Surg[Br] 1993;75-B:6-14.

79. Jeffrey R.McLaughlin, MD, and Kyla R. Lee, MD. Component Total Hip Arthroplasty with an Uncemented Tapered Femoral.J Bone Joint Surg Am. 2008;90:1290-1296.

ANNEXURE

Figure 11 : Immediate post operative and follow up x-ray showing penetration of femoral head into the acetabular cup indicating excessive PE liner wear. Also osteolysis seen in Gruen zones1, 2 and 7.

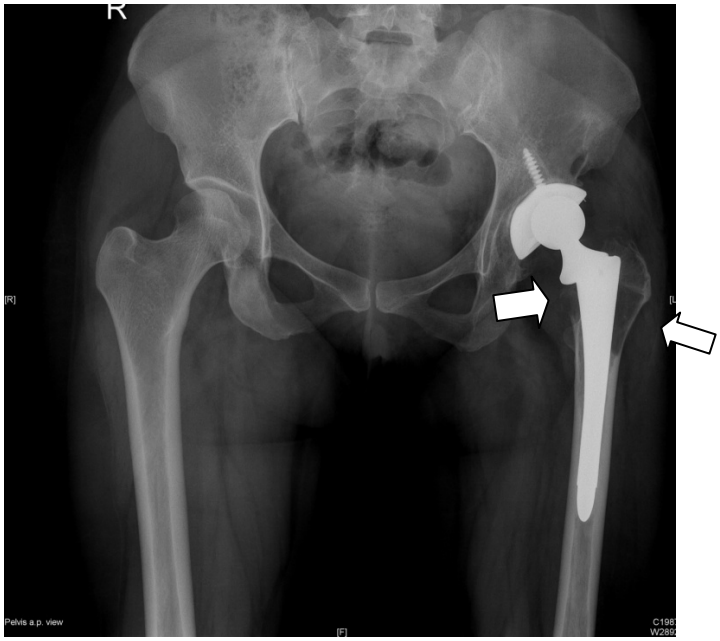

Figure 12 : Immediate post operative and follow up radiograph showing proximal penetration of femoral head into the acetabulum indication PE wear

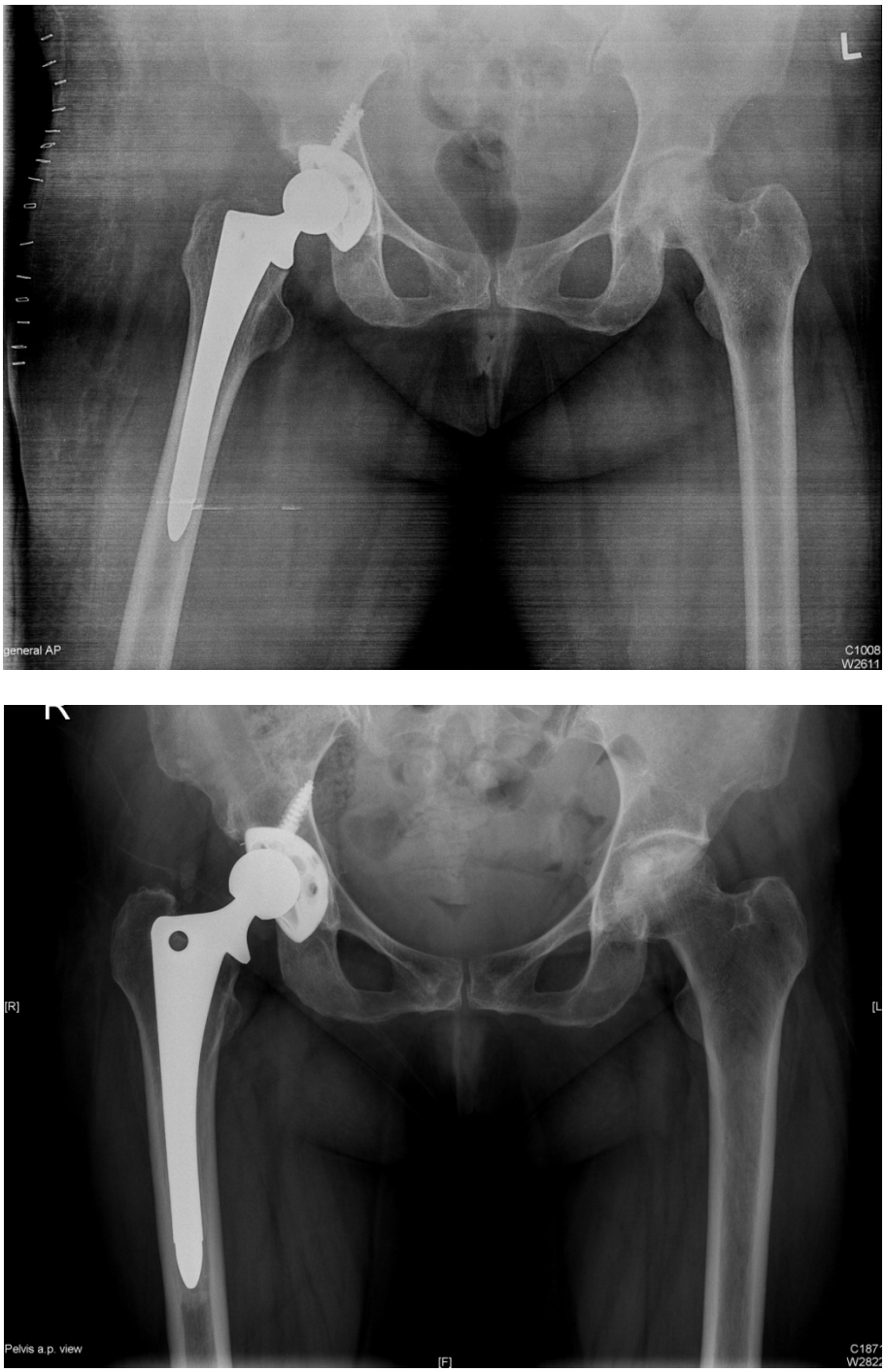

Figure 13: Right proximal femoral osteolysis at Gruen zones 1,2,6 and 7.

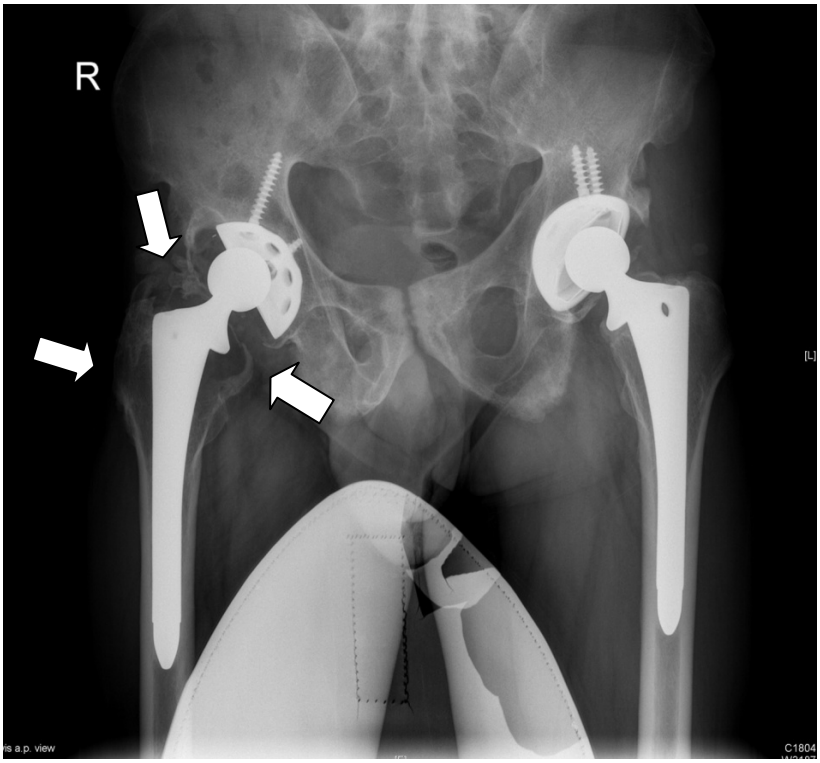

Figure 14: Left proximal femoral osteolysis involving Gruen zone 7

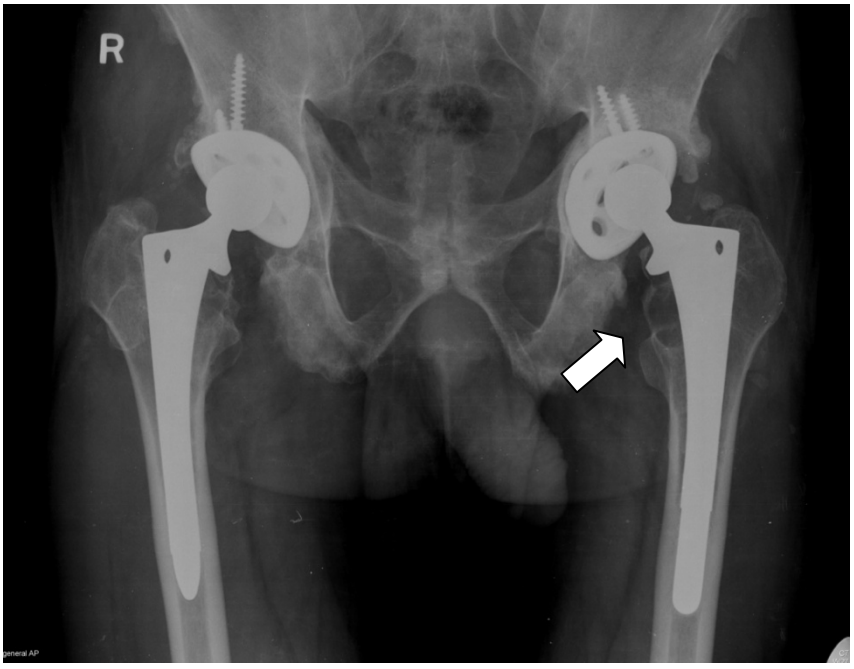

Figure 15: Immediate post operative and follow up radiograph. The lower radiograph shows proximal femur stress shielding and cortical hypertrophy at the tip of the stem.

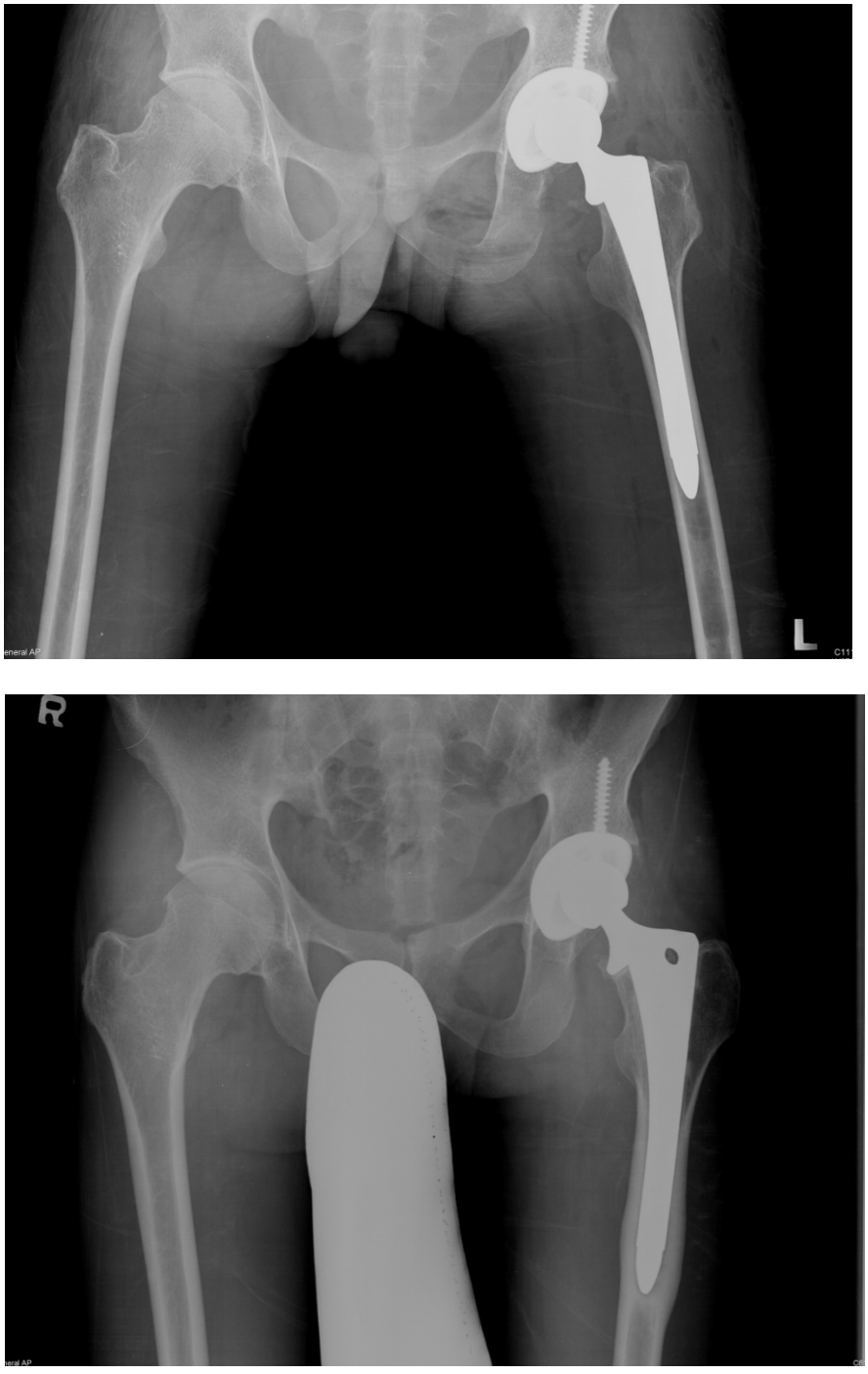

Figure 16: Right proximal femur stress shielding following uncemented THA with porous coated AML femoral stem.

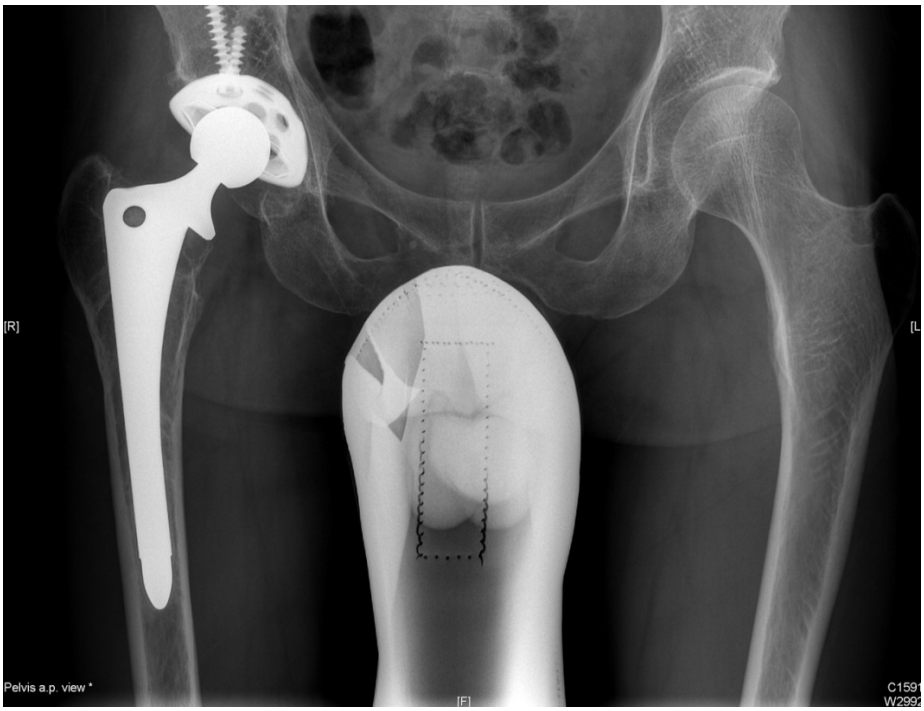

Table 2:Descriptives statistics for wear rate

Patient group		N	Minimum	Maximum	Mean	Std. Deviation
1	F/U Months	29	49	112	71.83	19.712
	AGE	29	18.00	61.00	37.4483	11.27129
	linear wear mm	29	.40	4.20	1.3000	.85315
	Wear rate mm/yr	29	.058	.525	.21290	.122770
	Vol wear rate mm3/yr	29	18.110	323.268	135.45810	79.053802
	Osteolysis Femur	29	0	2	.45	.686
	WITDH_MM	29	0	4	.69	1.538
	Migration	29	0	1	.03	.186
	Valid N (listwise)	29				
2	F/U Months	31	27	66	42.29	9.934
	AGE	31	16.00	67.00	42.9677	13.69059
	linear wear mm	31	.10	1.00	.3484	.25544
	Wear rate mm/yr	31	.020	.750	.11494	.139704
	Vol wear rate mm3/yr	31	13.683	205.250	57.65226	46.235801
	Osteolysis Femur	31	0	2	.10	.396
	WITDH_MM	31	0	0	.00	.000
	Migration	31	0	2	.13	.428
	Valid N (listwise)	31				

Table 3: NPar Tests :Mann-Whitney Test for wear rate

Ranks				
	Patient group	**N**	**Mean Rank**	**Sum of Ranks**
linear wear mm	**1**	29	43.78	1269.50
	2	31	18.08	560.50
	Total	60		
Wear rate mm/yr	**1**	29	40.67	1179.50
	2	31	20.98	650.50
	Total	60		
Vol wear rate mm3/yr	**1**	29	41.38	1200.00
	2	31	20.32	630.00
	Total	60		

Test Statistics(a)			
	linear wear mm	**Wear rate mm/yr**	**Vol wear rate mm3/yr**
Mann-Whitney U	64.500	154.500	134.000
Wilcoxon W	560.500	650.500	630.000
Z	-5.714	-4.369	-4.673
Asymp. Sig. (2-tailed)	.000	.000	.000
a Grouping Variable: Patient group			

Table 4:Chi Square test for acetabular osteolysis

Crosstab					
			Patient group		Total
			1	2	
Osteolysis Acetabulum	0	Count	24	31	55
		% within Patient group	82.8%	100.0%	91.7%
	1	Count	1		1
		% within Patient group	3.4%		1.7%
	2	Count	2		2
		% within Patient group	6.9%		3.3%
	3	Count	2		2
		% within Patient group	6.9%		3.3%
Total		Count	29	31	60
		% within Patient group	100.0%	100.0%	100.0%
Total		Count	29	31	60

	Value	df	Asymp. Sig. (2-sided)	Exact Sig. (2-sided)	Exact Sig. (1-sided)	Point Probability
Pearson Chi-Square	5.831(a)	3	.120	.022		
Likelihood Ratio	7.758	3	.051	.022		
Fisher's Exact Test	5.006			.022		
Linear-by-Linear Association	5.091(b)	1	.024	.022	.022	.022
N of Valid Cases	60					
a 6 cells (75.0%) have expected count less than 5. The minimum expected count is .48.						
b The standardized statistic is -2.256.						

Table 5: Chi Square test for femoral osteolysis

Crosstab					
			Patient group		Total
			1	2	
Osteolysis Femur	0	Count	19	29	48
		% within Patient group	65.5%	93.5%	80.0%
	1	Count	7	1	8
		% within Patient group	24.1%	3.2%	13.3%
	2	Count	3	1	4
		% within Patient group	10.3%	3.2%	6.7%
Total		Count	29	31	60
		% within Patient group	100.0%	100.0%	100.0%

	Value	df	Asymp. Sig. (2-sided)	Exact Sig. (2-sided)	Exact Sig. (1-sided)	Point Probability
Pearson Chi-Square	7.525(a)	2	.023	.017		
Likelihood Ratio	8.141	2	.017	.022		
Fisher's Exact Test	7.329			.017		
Linear-by-Linear Association	5.535(b)	1	.019	.022	.015	.011
N of Valid Cases	60					

a 4 cells (66.7%) have expected count less than 5. The minimum expected count is 1.93.

b The standardized statistic is -2.353.

Table 6: Chi Square test for stress shielding

Crosstab					
			Patient group		Total
			1	2	
Femur stress shielding	0	Count	12	13	25
		% within Patient group	41.4%	41.9%	41.7%
	1	Count	8	6	14
		% within Patient group	27.6%	19.4%	23.3%
	2	Count	8	8	16
		% within Patient group	27.6%	25.8%	26.7%
	3	Count		3	3
		% within Patient group		9.7%	5.0%
	4	Count	1	1	2
		% within Patient group	3.4%	3.2%	3.3%
Total		Count	29	31	60
		% within Patient group	100.0%	100.0%	100.0%

Chi-Square Tests						
	Value	df	Asymp. Sig. (2-sided)	Exact Sig. (2-sided)	Exact Sig. (1-sided)	Point Probability
Pearson Chi-Square	3.263(a)	4	.515	.579		
Likelihood Ratio	4.419	4	.352	.495		
Fisher's Exact Test	3.173			.583		
Linear-by-Linear Association	.334(b)	1	.564	.641	.325	.079
N of Valid Cases	60					
a 4 cells (40.0%) have expected count less than 5. The minimum expected count is .97.						
b The standardized statistic is .578.						

Table 7: p value for Harris Hip Score

Levene's Test for Equality of Variances		t-test for Equality of Means						
F	Sig.	t	df	Sig. (2-tailed)	Mean Difference	Std. Error Difference	95% Confidence Interval of the Difference	
							Lower	Upper
16.330	.000	1.548	27	.133	2.68	1.734	-.874	6.243
		1.773	25.774	.088	2.68	1.514	-.430	5.798
10.577	.003	-1.321	27	.198	-.63	.478	-1.612	.349
		-1.837	18.000	.083	-.63	.344	-1.354	.091
6.368	.018	1.098	27	.282	2.55	2.319	-2.211	7.306
		1.342	26.931	.191	2.55	1.898	-1.348	6.443
.013	.909	.419	27	.679	.66	1.584	-2.588	3.914
		.414	17.899	.684	.66	1.601	-2.701	4.028
.659	.424	.773	27	.446	2.33	3.010	-3.849	8.502
		.843	23.275	.408	2.33	2.759	-3.378	8.030
.077	.783	.077	27	.940	.13	1.649	-3.258	3.510
		.076	17.818	.941	.13	1.669	-3.383	3.635

PROFORMA

Name:
Hospital number:
Age/sex:
Height:
Weight:
BMI:
Permanent full residential address:

Phone no:
Email ID:

Date of surgery:
Last review:
Duration of follow up:

Side - left/right/BL:
Smoking- Y/N
Comorbidity - DM/HTN/IHD/Asthma/others

Diagnosis:

Charnley class: A B C

Functional score: Harris Hip Score
Pre op score
Post op score:

Wear rate: (Post op xray with last follow up xray)
Computer assisted:

Total linear wear:
Linear wear rate:
Volumetric wear rate:

Osteolysis:

Acetabulum: present absent
Type 1
Type2
Type3

Witdh 1) <0.5cm
2)<1cm
3)<1.5cm
4)>1.5cm

Femur: present absent

Gruen: 1 2 3 4 5 6 7

Grade: mild(<2) extensive(2-5) extensive(>6)

Migration: 1) subsidence >2mm

2)varur/valgus tilt >5°

Stress shielding: present absent

Degree 1° 2° 3° 4°

Harris Hip Score

Hip ID:
Study Hip: ☐ Left ☐ Right
Examination Date (MM/DD/YY): / /
Subject Initials: |____|____|____|
Medical Record Number:

Interval: _____

Harris Hip Score

Pain *(check one)*

- ☐ None or ignores it (44)
- ☐ Slight, occasional, no compromise in activities (40)
- ☐ Mild pain, no effect on average activities, rarely moderate pain with unusual activity; may take aspirin (30)
- ☐ Moderate Pain, tolerable but makes concession to pain. Some limitation of ordinary activity or work. May require Occasional pain medication stronger than aspirin (20)
- ☐ Marked pain, serious limitation of activities (10)
- ☐ Totally disabled, crippled, pain in bed, bedridden (0)

Limp

- ☐ None (11)
- ☐ Slight (8)
- ☐ Moderate (5)
- ☐ Severe (0)

Support

- ☐ None (11)
- ☐ Cane for long walks (7)
- ☐ Cane most of time (5)
- ☐ One crutch (3)
- ☐ Two canes (2)
- ☐ Two crutches or not able to walk (0)

Distance Walked

- ☐ Unlimited (11)
- ☐ Six blocks (8)
- ☐ Two or three blocks (5)
- ☐ Indoors only (2)
- ☐ Bed and chair only (0)

Sitting

- ☐ Comfortably in ordinary chair for one hour (5)
- ☐ On a high chair for 30 minutes (3)
- ☐ Unable to sit comfortably in any chair (0)

Enter public transportation

- ☐ Yes (1)
- ☐ No (0)

Stairs

- ☐ Normally without using a railing (4)
- ☐ Normally using a railing (2)
- ☐ In any manner (1)
- ☐ Unable to do stairs (0)

Put on Shoes and Socks

- ☐ With ease (4)
- ☐ With difficulty (2)
- ☐ Unable (0)

Absence of Deformity (All yes = 4; Less than 4 =0)

- Less than 30° fixed flexion contracture ☐ Yes ☐ No
- Less than 10° fixed abduction ☐ Yes ☐ No
- Less than 10° fixed internal rotation in extension ☐ Yes ☐ No
- Limb length discrepancy less than 3.2 cm ☐ Yes ☐ No

Range of Motion (**indicates normal*)

- Flexion (*140°) _____
- Abduction (*40°) _____
- Adduction (*40°) _____
- External Rotation (*40°) _____
- Internal Rotation (*40°) _____

Range of Motion Scale

211° - 300° (5)	81° - 100 (2)
161° - 210° (4)	31° - 80° (1)
101° - 160° (3)	0° - 30° (0)

Range of Motion Score _____

Total Harris Hip Score _____

Enduron

S. N	Hospital no	Age	Sex	Side	Diagnosis	F/U Months	linear wear mm	Wear rate mm/yr	Vol wear rate mm3/yr	angle of penetration	Osteolysis Acetabulum	width mm	Migration	Osteolysis Femur	Femur stress shielding	Charnley class
1	931245B	60	1	2	3	103	1.3	0.152	94.174	3	0	0	0	0	2	A
2	888308B	28	1	1	8	75	1.5	0.25	153.937	2	0	0	0	0	2	B
3	888308B 1	28	1	2	8	75	1.1	0.183	153.937	1	3	4	0	0	2	B
4	816763B	39	1	1	1	60	1.1	0.22	135.465	1	0	0	0	0	1	B
5	816763B 1	39	1		1	60	1.2	0.24	147.78	1	0	0	0	0	0	B
6	956718B	30	1	1	1	108	2.6	0.28	177.883	2	0	0	0	2	0	B
7	956718B 1	30	1	2	1	71	0.7	0.116	71.837	3	0	0	0	0	0	B
8	946516B	42	1	1	3	112	1.6	0.177	109.466	2	0	0	0	2	2	A
9	793687B	61	1	1	3	111	0.5	0.058	18.11	3	2	4	1	1	0	A
10	823246B	43	2	2	7	70	1.1	0.183	112.87	1	0	0	0	1	1	A
11	676424A	35	2	1	4	83	2.3	0.32	202.317	1	0	0	0	0	0	C
12	129917C	22	1	1	1	78	1.5	0.23	142.096	1	0	0	0	1	4	B
13	995211B	42	1	1	3	56	1	0.2	123.15	1	0	0	0	0	2	B
14	995211B 1	42	1	2	3	56	0.8	0.16	98.52	1	0	0	0	0	1	B
15	763330B	18	1	2	1	61	0.7	0.14	86.205	1	0	0	0	0	0	B
16	065472C	56	1	2	3	80	0.6	0.085	52.778	1	0	0	0	0	1	A
17	072742C	30	2	2	9	107	4.2	0.525	323.268	1	0	0	0	2	2	A
18	593715B	35	1	1	1	57	2	0.4	246.3	1	0	0	0	1	2	B
19	593715B	35	1	2	1	57	2.2	0.44	270.93	1	0	0	0	1	1	B
20	056370C	26	1	2	8	58	2.4	0.48	296.56	1	0	0	0	1	1	B
21	117883C	45	1	2	1	57	0.6	0.133	82.1	1	0	0	0	0	0	B
22	124965C	35	1	1	4	49	0.5	0.125	76.968	3	3	4	0	0	0	B
23	124965C 1	35	1	2	4	49	0.4	0.1	61.575	1	2	4	0	1	0	B
24	124878C	27	1	2	5	55	1.4	0.311	191.566	1	0	0	0	0	2	A
25	004871C	51	2	1	3	53	0.5	0.111	68.416	1	0	0	0	0	0	A
26	030893C	41	1	2	10	86	0.7	0.1	61.575	1	0	0	0	0	1	A
27	077593C	56	1	2	1	60	0.4	0.08	49.26	1	1	4	0	0	0	B
28	250058C	24	1	1	1	77	0.9	0.138	85.257	1	0	0	0	0	0	B
29	315584C	31	1	1	4	59	1.9	0.237	233.985	1	0	0	0	0	1	A

Marathon

S. N	Hospital no	Age	Sex	Side	Diagnosis	F/u in months	Linear wear mm	Wear rate mm/yr	Vol wear rate mm3/yr	angle of penetration	Oeteolysis acetabulum	Width mm	Migration	Osteolysis Femur	Femur stress shielding	Charnley class
1	341064C	31	1	2	1	42	0.4	0.088	54.733	2	0	0	0	0	0	B
2	381311C	50	1	2	6	45	0.7	0.175	107.756	2	0	0	0	0	1	A
3	394393C	38	1	1	3	60	0.4	0.08	49.26	2	0	0	0	0	1	A
4	278270C	21	2	2	11	66	0.8	0.145	89.563	1	0	0	0	0	0	B
5	278270C	21	2	1	11	46	0.7	0.175	107.756	1	0	0	0	0	0	B
6	315051C	24	1	1	12	57	0.5	0.1	61.575	1	0	0	0	0	3	A
7	428849C	46	1	1	13	41	0.4	0.114	70.371	1	0	0	0	0	3	A
8	491582C	49	1	2	3	50	0.5	0.119	76.968	2	0	0	0	0	1	B
9	490526C	34	1	1	1	60	0.4	0.08	49.26	1	0	0	0	2	2	B
10	271673C	54	2	2	5	44	0.3	0.075	46.181	1	0	0	0	0	0	B
11	493068B	16	1	2	1	33	0.3	0.1	61.575	2	0	0	0	0	0	B
12	795962B	45	1	2	1	49	0.3	0.075	46.181	1	0	0	0	0	0	B
13	486306C	41	1	1	4	50	0.2	0.05	30.787	1	0	0	0	0	2	A
14	072300A	67	1	2	14	44	0.2	0.05	30.787	1	0	0	0	0	0	C
15	455315B	26	1	2	1	51	0.1	0.02	13.683	1	0	0	1	0	1	B
16	549407C	49	1	1	1	29	0.2	0.066	41.05	2	0	0	0	0	1	C
17	633245C	65	1	2	3	43	0.2	0.05	30.787	2	0	0	0	1	1	C
18	648839C	55	1	2	3	47	0.3	0.075	46.181	1	0	0	0	0	2	C
19	191762B	59	2	2	9	37	0.1	0.033	20.525	1	0	0	0	0	2	A
20	696104C	49	1	1	15	45	0.3	0.75	46.181	1	0	0	2	0	4	A
21	736485B	47	1	2	4	39	0.4	0.1	61.575	1	0	0	0	0	3	A
22	810074C	51	1	2	4	30	0.2	0.08	49.26	2	0	0	1	0	0	A
23	344445C	40	1	1	1	33	0.1	0.033	20.525	1	0	0	0	0	0	B
24	344445C 1	40	1	2	1	27	0.1	0.033	20.525	1	0	0	0	0	0	B
25	238787	65	1	2	14	34	0.2	0.066	41.05	1	0	0	0	0	0	C
26	964412B	36	1	1	1	37	1	0.333	205.25	2	0	0	0	0	2	B
27	964412B 1	36	1	2	1	37	0.1	0.033	20.525	1	0	0	0	0	2	B

Figure 1: Basic components in a Total Hip Arthroplasty

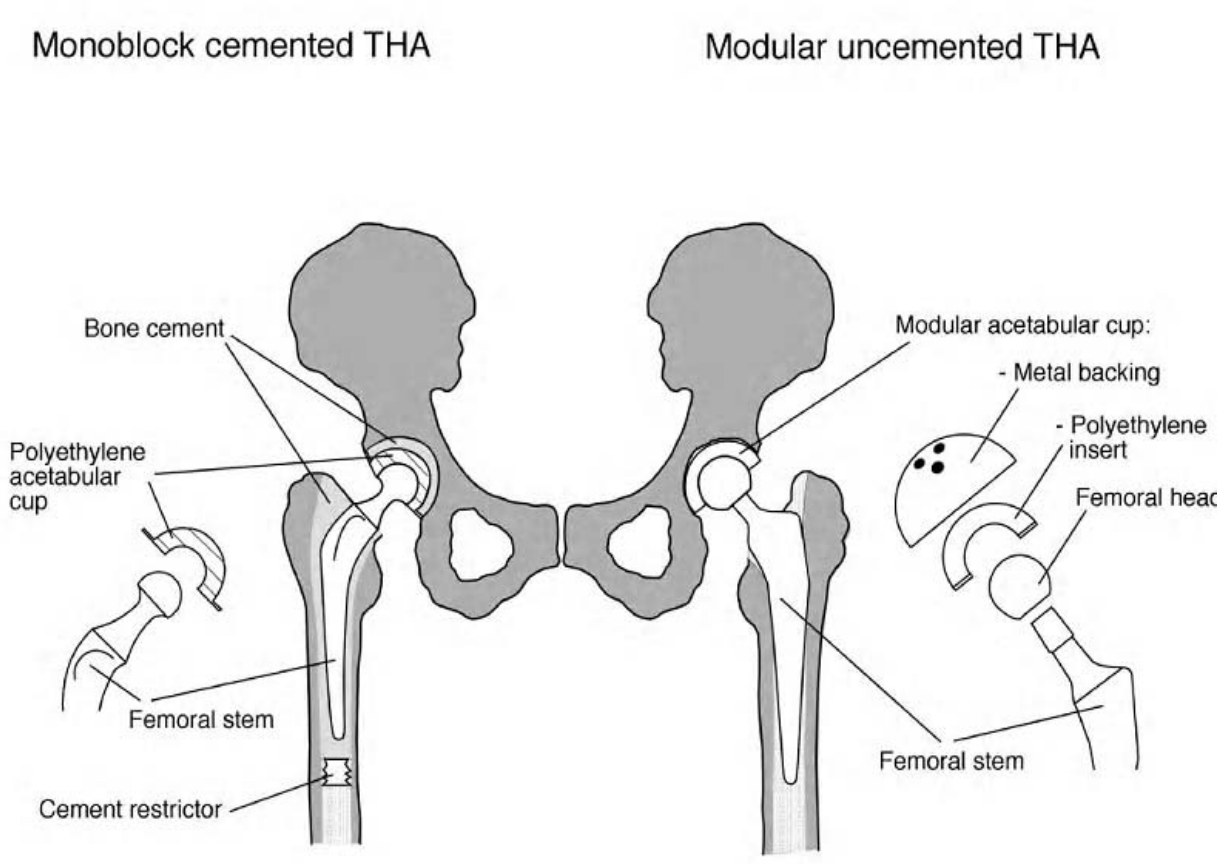

Table1. Patient demographics

	Conventional PE	**X-linked PE**
Total no. of hips	29	27
Total no. of patients	23	24
Bilateral arthroplasty	6 (26.1%)	3 (12.5%)
Right [n (%)]	13(44.82%)	10 (37.03%)
Left [n (%)]	16 (55.18%)	17 (62.97%)
Male [n (%)]	19 (82.6%)	21 (87.5%)
Female [n (%)]	4 (17.4%)	3(12.5%)
Mean age [y (range)]	37.45 (18-61)	42.97 (16-67)

Figure 2: Comparing linear wear rate

	Conventional PE	X-linked PE
Wear rate mm/yr	0.21	0.11

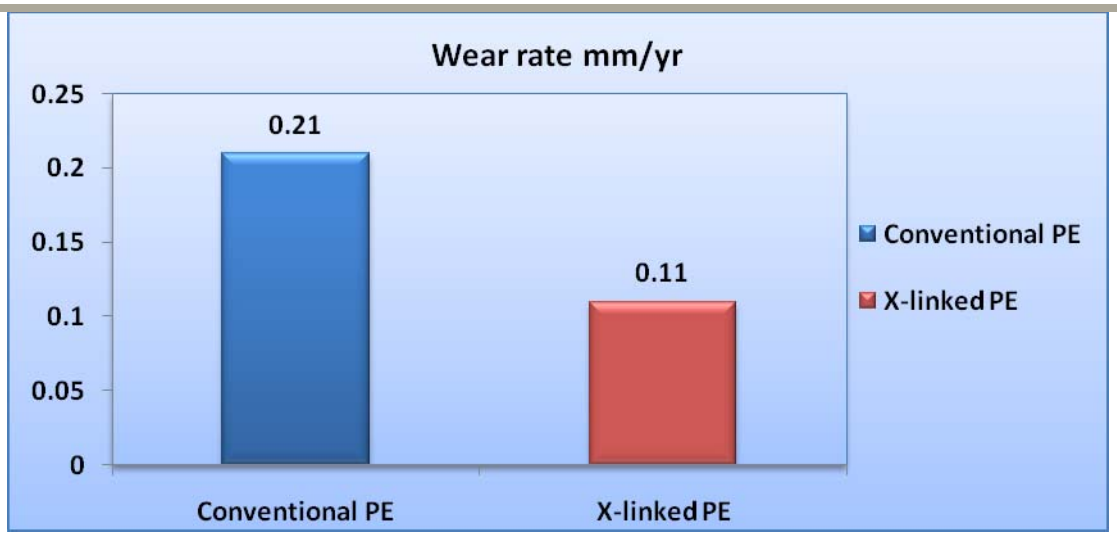

Figure 3: Comparing volumetric wear rate

	Conventional PE	X-linked PE
Volumetric wear rate mm^3/yr	135.46	55.55

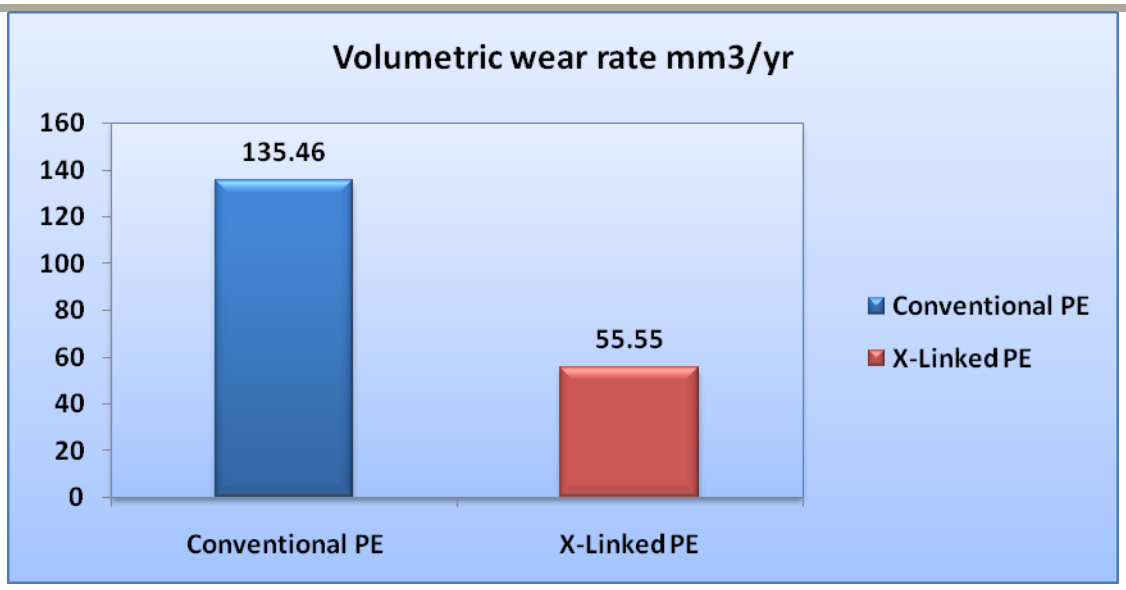

Figure 4:Angle of penetration (direction of wear) in the Conventional PE group

	n (neutral)	θ (medial)	α (lateral)
Angle of penetration	22	3	4

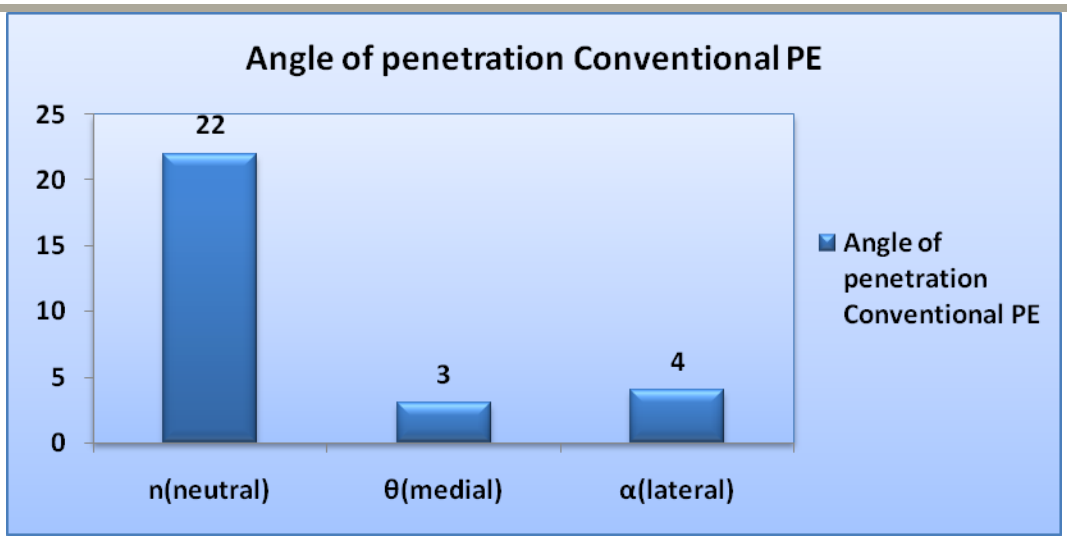

Figure 5:Angle of penetration (direction of wear) in the X-linked PE group

	n (neutral)	θ (medial)	α (lateral)
Angle of penetration	18	9	0

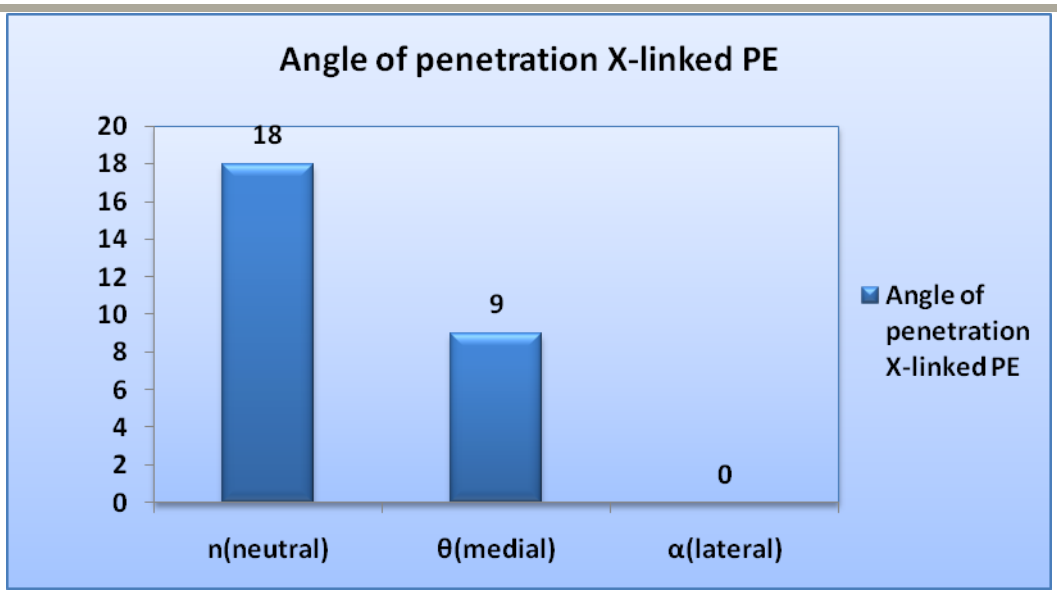

Figure 6: Comparison of acetabular osteolysis

	Absent	Type 1	Type 2	Type 3
Conventional PE	24	1	2	2
X-linked PE	27	0	0	0

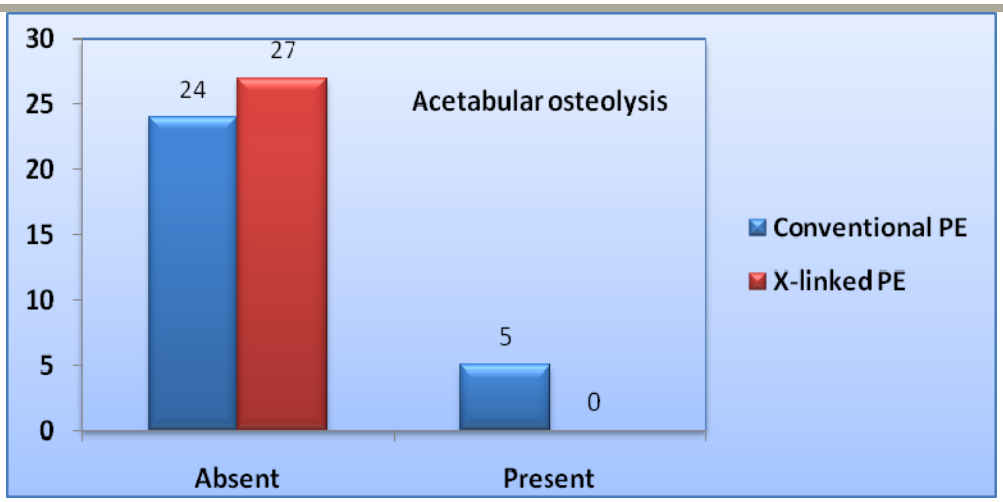

Figure 7:Comparison of femoral osteolysis

	none	mild	intermediate	extensive
Conventional PE	19	7	3	0
X-linked PE	25		1	0

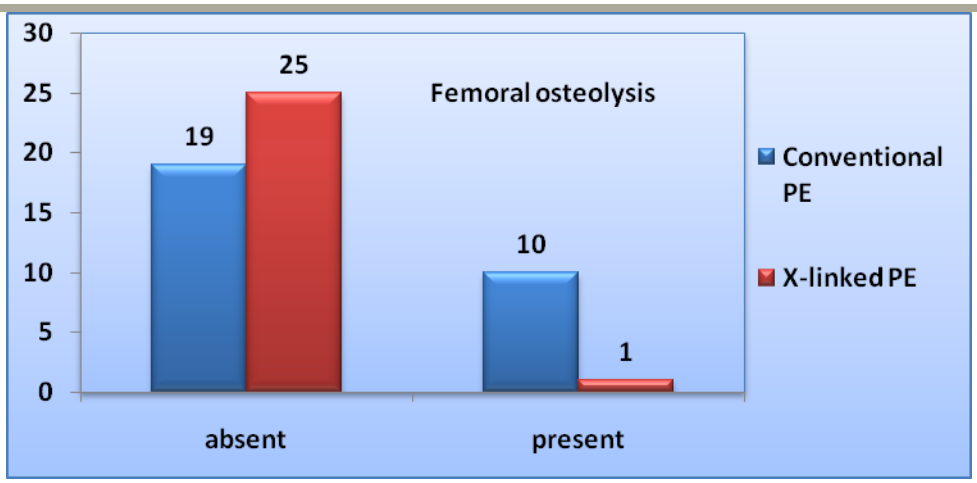

Figure 8:Comparison of stress shielding

	none	1st degree	2nd degree	3rd degree	4th degree
Conventional PE	12	8	8	0	1
X-linked PE	11	6	6	3	1

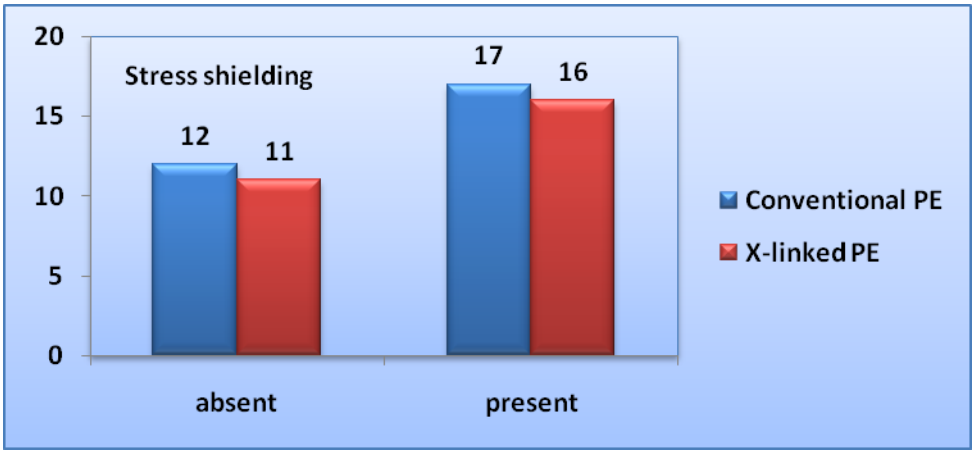

Figure 9:Comparison of Harris Hip Score

	Pain score pre op	Pain score post op	Functional score pre op	Functional score post op	Harris hip score pre op	Harris hip score post op
Conventional PE	19	40	20.6	37.4	40.8	85.6
X-linked PE	16.31	40.63	18.05	36.73	38.47	85.47

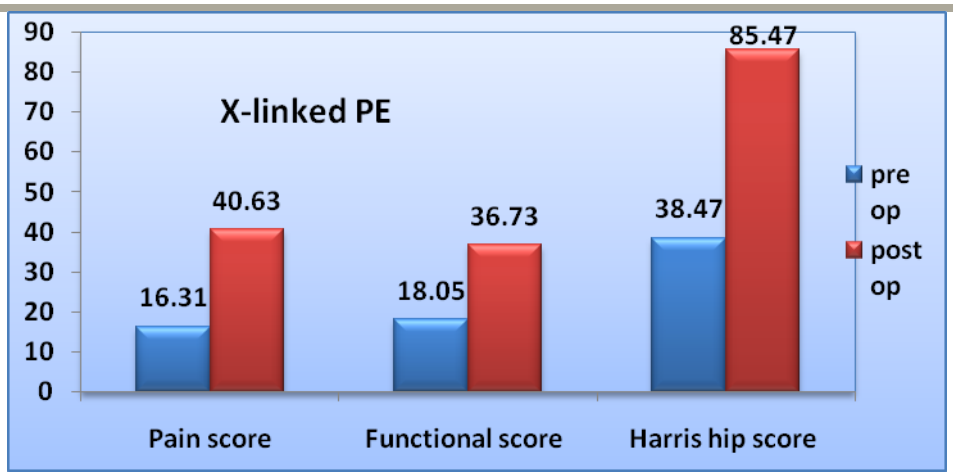

Figure 10: Comparison of wear rate in patients with and without osteolysis

Wear rate mm/yr	0-0.09	0.1-0.19	0.2-0.29	>0.3
With osteolysis	4	5	2	4
Without osteolysis	18	15	5	3